Magnetically Actuated Soft Miniature Robots for Applications in the Urinary Tract

by

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Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

Abstract

Kidney stones, affecting approximately 10% of the global population, pose significant health concerns due to their prevalence and recurrence. The formation of these stones, known medically as nephrolithiasis, involves the aggregation of various types of crystals such as calcium oxalate, uric acid, struvite, and cystine. Traditional treatments range from pain management for mild cases to invasive procedures for severe obstructions. However, the high recurrence rates of the disease and the complications of current treatments necessitate innovative and minimally invasive solutions.

This thesis explores the development of a small-scale soft magnetic robot designed to facilitate the dissolution of kidney stones and prevent their recurrence. The robot uses magnetic actuation, a preferred method due to its minimal interaction with human tissues and reliable control.

The magnets used for actuation are modeled in MATLAB and COMSOL Multiphysics to observe and compare the fields in different sizes and distances, furthermore, the forces and torques are calculated for different cases. After modeling the actuation, filamentous magnetic robots made of Gelatin-methacrylate were designed and experimented with in 3D-printed urinary tract organ models which showed their maneuverability in their target environment, with analysis done on the best location and orientation of the magnet inside the filament. Furthermore, the filaments were loaded with different drug choices to determine an efficient chemical for the dissolution of uric acid kidney stones.

We highlight the integration of miniature robots in medical applications, emphasizing their potential for targeted drug delivery, minimally invasive procedures, and real-time diagnostics.

This research shows that the robot configuration which has the actuation magnet perpendicular to the robot magnet has the capacity for movements up to 3 times faster than parallel-placed magnets, their movement reaching about 18 mm/s. However, this is only true in confined spaces and in non-confined environments, parallel-placed magnets in the robot and actuator show stability and reliability with speeds of 8 mm/s. Experiments showed no significance in the location of the robot magnet placement along the filament

and the addition of the active chemical to the filaments showed a mass reduction of about 30% in uric acid stones, which is double the amount of control samples.

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I express my heartfelt gratitude to everyone who contributed to the completion of this thesis. This work would not have been achievable without the invaluable supervision of Dr. Veronika Magndanz, whose supervision was instrumental in guiding me toward conducting high-quality research for enhancing medical procedures. My sincere thanks also go to Dr. Behrad Khamesee for sharing his expertise in the field of magnetics.

I am deeply appreciative of all the members of the medical microrobots lab, including Aaron Lewis, Erica Liu, Megheti Bekmezian, Mathilda LaBrash-White, Noura Abdelrazec, Lauren Kwong, and the members of the MagLev lab. Additionally, I am grateful to Chris McClellan from the Department of Systems Design Engineering for his technical support.

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Finally, I acknowledge the essential support of the Graham Seed Fund, whose assistance was crucial to the realization of this project.

Dedication

This is dedicated to all my dear ones, my supportive parents and my loving partner.

I would like to extend my heartfelt thanks to my best friends, K.P. and E.K., who have stood by me since high school; to all my friends at UW and to Sam Vandekerckhove, this thesis is for you, for your unwavering support and encouragement that made this possible.

Table of Contents

Author's Declaration ii

Abstract iii

Acknowledgements v

Dedication vi

List of Figures x

List of Tables xv

1 Introduction 1

1.1 Kidney Stone Disease 1

1.2 Miniature Robots for Medical Applications 3

1.3 Magnetic Actuation 5

1.4 Objectives and Contributions 7

1.5 Thesis Outline 8

2 Literature Review 10

2.1 Magnetic Actuation: Electromagnets 10

2.2 Magnetic Actuation: Permanent Magnets 16
2.3 Magnetic robot design and fabrication 20
2.4 Medical Applications of Magnetic Robots 27

3 Methodology **30**
3.1 Magnetic Theory 31
3.2 Mechanical Properties and Hydrodynamics 41
3.3 Actuation System Overview 45
3.3.1 Robotic Arm 45
3.3.2 Rotating Magnetic Field 47
3.3.3 Camera and Tracking 47
3.3.4 Anatomical Model of the Urinary Tract 48
3.4 Robot Fabrication and Kidney Stone Dissolution 49
3.5 Experimental Design 52
3.5.1 Locomotion and frequency response 52
3.5.2 Dissolution of Kidney Stones 55

4 Experimental Results and Discussion **57**
4.1 Locomotion and Frequency Response 57
4.2 Filament Deformation 67
4.3 Kidney Stone Dissolution 68

5 Conclusions and Future Work **73**

6 Letters of Copyright Permission **78**

References **152**

APPENDICES **159**

A **160**

A.1 Arduino Code for Controlling and Monitoring the Actuation Frequency . . 160

A.2 MATLAB Code for Modelling the Field 163

List of Figures

1.1 Schematic of the human kidney with a stone in the renal pelvis [3] 2

1.2 (a) Magnetic field lines produced by a coil [11] (© 2020 Springer Nature) (b) Magnetic field lines produced by a permanent magnet [12] (© 2018 Springer Nature) 5

2.1 (a) Helmholtz coil pair (b) Maxwell coil pair (c) Set-up used in [15] consisting of 3 Helmholtz and 2 Maxwell pairs (Reprinted from the cited source © 2010, with permission from Elsevier) (d) Set-up used in [16] consisting of one Maxwell coil pair, one Helmholtz coil pair and one saddle coil (© 2010 IEEE) 11

2.2 (a) Octomag [18] (© 2010 IEEE) (b) Rectamag3D [19] (© 2020 IEEE) . . 13

2.3 (a) ARMM [20] (© 2023 IEEE) (b) DEMA [21] (Reprinted from the cited source © 2023, with permission from Elsevier) (c) RAMAN [22] (© 2024 IEEE) 14

2.4 (a) Single magnet to push and pull [24] (b) Multiple rotating magnets [27] (© 2017 IEEE) (c) The Stereotaxis Niobe Robotic Magnetic Navigation System [28] (© 2007 John Wiley and Sons) (d) Single rotating magnet [25] (© 2014 IEEE) 19

2.5 (a) Oscillatory fin on a microrobot [17] (© 2006 Sage Publications) (b) Helical motion in a tube 1.[33] 2.[38] (© 2010 IEEE) (c) Rolling motion [39] (© 2010 AIP Publishing) 22

2.6 The rod-shaped magnetic robot design of Diller *et al.* (a) [42] (© 2014 AIP Publishing) and Hu *et al.* (b) [43] (© 2018 Springer Nature), and their deformation in different magnetic fields. 23

2.7 (a) Ellipsoid magnetic microrobot [45] (© 2006 Sage Publications) (b) Fish-like magnetic microrobot [46] (© 2012 IOP Publishing) (c) Flower-like soft microrobot [47] (Reprinted from the cited source © 2016, American Chemical Society) 25

2.8 (a) Endoscopic capsule [24] (b) Hygdogel spherical robot for intravascular drug delivery [48] (© 2023 Royal Society of Chemistry) (c) Magnetic optical oxygen sensor [49] (© 2008 IEEE) (d) Rough spherical magnetic miniature robot for drilling and removing blood clots [15] (Reprinted from the cited source © 2010, with permission from Elsevier) (e) Soft gripper robot [50] (© 2017 IEEE) (f) Magnetic microrobot for stem cell culture and delivery [51] (© 2013 John Wiley and Sons) 28

3.1 (a) Robot arm from igus (b) DC motor attached to the end-effector of robot arm (c) Arduino Uno as microcontroller to monitor the RPM (d) Permanent magnet inside a holder rotated by the DC motor (e) Proposed configurations for the soft filament miniature robots 31

3.2 (a) The actuator magnet and the robot magnet are perpendicular, making the fin-like Robots. (b) The actuator magnet and the robot magnet are parallel to each other making the screw-like Robots. 32

3.3 Optimization of the Radius of the magnet for a height of 25 mm 34

3.4 (a) Actuator magnet, (b) Robot magnet 35

3.5 (a) The magnetic field produced by the cylindrical actuator magnet along its center-line (b) Torque (Nm) induced on fin-like robot at t=0s over distance from actuator magnet(c) Gradient of the magnetic field produced by the cylindrical actuator magnet along its center-line (d) Force (N) induced on screw-like robot at t=0s over distance from actuator magnet 38

3.6 Comparison of the rotation of the robot vs. the rotation of the actuator magnet 41

3.7 (a) Deflection of a beam due to applied moment (b) Deflection of a beam due to applied point force 42

3.8 The 5 cases of magnet placement studied in this work for frequency response and deformation for both the fin-like and the screw-like robots. The head of the robot is indicated by the black square/ robot magnet in this figure. Assuming the swimming direction for all five cases to be to the left, cases 1 and 2 have the head of the robot moving in that direction and cases 4 and 5 have the tail moving first. 43

3.9 (a) Deformation of fin-like robots due to applied moment (b) Deformation of screw-like robots due to applied point force 44

3.10 Robot Arm from igus 46

3.11 (a) DC motor attached to the robot arm's end-effector, (b) Actuator magnet inside a holder 48

3.12 (a) Design of the PDMS mold (b) PDMS mold (c) Dyed GelMa filament with embedded permanent magnet (d) GelMa filament and Kidney stone after dissolution experiment (e) The chemical reaction catalysed by urease through which urine becomes basic, in this reaction urea reacts with water resulting in carbon dioxide and ammonia (Molecule figures taken from PubChem) 50

3.13 Filaments in the ureter model in a tube of water indicated in , placed under actuation and being recorded 53

3.14 Schematic of the human urinary tract [61] (© 2023 Springer Nature) (a) and the full 3D-printed organ model used for actuation experiments in this research (b) 54

4.1 (a) Movement of the fin-like robots inside a tubular environment with the magnet embedded as case 1 over a time period of 6 (s), (b) Movement of the screw-like robots inside a tubular environment with the magnet embedded as case 1 over a time period of 6 (s) 58

4.2 (a) Movement of the fin-like robots inside a tubular environment with the magnet embedded as case 2 over a time period of 6 (s), (b) Movement of the screw-like robots inside a tubular environment with the magnet embedded as case 2 over a time period of 6 (s) 58

4.3 (a) Movement of the fin-like robots inside a tubular environment with the magnet embedded as case 3 over a time period of 6 (s), (b) Movement of the screw-like robots inside a tubular environment with the magnet embedded as case 3 over a time period of 6 (s) 59

4.4 (a) Movement of the fin-like robots inside a tubular environment with the magnet embedded as case 4 over a time period of 6 (s), (b) Movement of the screw-like robots inside a tubular environment with the magnet embedded as case 4 over a time period of 6 (s) 59

4.5 (a) Movement of the fin-like robots inside a tubular environment with the magnet embedded as case 5 over a time period of 6 (s), (b) Movement of the screw-like robots inside a tubular environment with the magnet embedded as case 5 over a time period of 6 (s) 60

4.6 (a) Movement of the screw-like robots inside a tubular environment at 16 Hz , (b) Movement of the fin-like robots inside a tubular environment at 16 Hz 60

4.7 Frequency response of the screw-like (a) and the fin-like (b) robots 61

4.8 Frequency response of the fin-like (a) and the screw-like (b) robots for case 2 and 4 combined at higher frequencies 62

4.9 Guidance of the fin-like (a) and the screw-like (b) robots in the bladder at 8Hz, the red box indicates the location of the robot 63

4.10 Frequency response of the screw-like robots in the bladder 64

4.11 Guidance of the screw-like (a) and the fin-like (b) robots in the renal pelvis at 8Hz, the red box indicates the location of the robot 65

4.12 (a) Modelling of the deformation of the fin-like robots due to the applied moment (b) Experimental deformation of the fin-like robots due to the applied moment (c) Modelling of the deformation of the screw-like robots due to the applied point force (b) Experimental deformation of the screw-like robots due to the applied point force 68

4.13 (a) Effectiveness of different concentrations of urease embedded in filaments over 24 hours, the increase in urine pH indicates the efficacy of urease (b) Phenol red pH indicator showing the immediate effect of urease in increasing urine pH, arrow shows the filament containing a micromagnet and urease (c) Effectiveness of urease in increasing urine pH in long term, where the filaments were loaded with different concentrations of urease and placed in synthetic urine with pH 6 and the solution was changed every 14 hours . . 69

4.14 (a) Filament and stone after dissolution experiment with crystals formed on the stone (b) Scatter plot showing the most effective dissolution of stones at different pH levels. the percentage of mass reduction for stones in contact with different concentrations of urease is plotted against the average pH after replacement to fresh urine of pH 5.5 every 24h over 4 days (c) Dissolution of kidney stones due to the presence of urease where the average percentage of mass reduction of uric acid stones is compared to different amounts of urease used, showing the amount of 5mg/mL of urease to be the most effective 71

List of Tables

2.1 Comparison of Electromagnets and Permanent Magnets 18

3.1 Symbolic Variables, Descriptions, and Unknowns 40

Chapter 1

Introduction

1.1 Kidney Stone Disease

Kidneys are one of the most crucial organs in the human body, and kidney stones are a common problem occurring worldwide and causing significant health concerns. They occur globally, affecting 10% of the people worldwide [1]. Kidney stones, medically known as nephrolithiasis or renal calculi, develop when certain chemicals become concentrated in kidneys, forming different types of crystals such as calcium oxalate, uric acid, struvite, and cystine. Each type of stone has distinct properties and may require specific treatments. The issue is becoming more and more prevalent and incident due to the sedentary lifestyle of the human population, obesity, and unhealthy diet. Renal calculi are more prevalent in males [2], in addition, once a person suffers from a kidney stone, there is a 70-80 % chance of recurring stones [2].

Kidney stones may cause acute pain, inflammation and infection. Furthermore, if a stone is lodged in the ureter, and obstructs the flow of urine, it leads to a condition known as urinary retention which further increases the risk of urinary tract infections (UTIs) and chronic kidney disease. Modern medicine offers various treatments for different cases of nephrolithiasis. In mild cases where the stone is able to pass through the urinary tract, the treatment is mainly pain relief medication. However, in more severe cases,

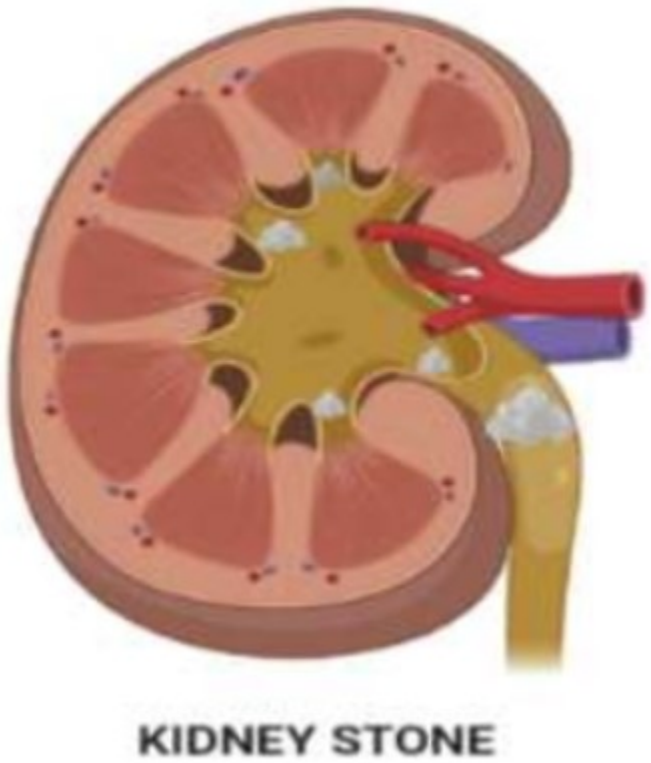


Figure 1.1: Schematic of the human kidney with a stone in the renal pelvis [3]

interventions including oral medication, shockwave therapy, ureteroscopy and minimally invasive surgeries are needed.

Uric acid stones form when the urine becomes too acidic due to the presence of uric acid. For the specific case of uric acid stones, there is a treatment called chemolysis (alkalization of urine) which could be administered orally. By administration of medications containing bicarbonate or citrate, the pH of the urine increases to higher than 6, which is known to help the chemolysis [1]. However, the oral administration of these medications may not be ideal as it may cause digestion issues or the digestive chemical reactions may interfere. Furthermore, the treatment needs to be applied for elongated periods of time with several

doses per day [4]. Therefore, targeted drug delivery of a substance which triggers chemolysis would be the preferred method.

Innovative methods have been suggested for breaking down uric acid kidney stones by using basifying agents like N-acetylcysteine or theobromine [1]. These agents show potential for use in oral treatments, complementing the effectiveness of citrate or bicarbonate. These agents are occasionally used by inserting a catheter and initiating localized chemolysis. However, catheter-based chemolysis includes the risk of infection or sepsis due to continuous irrigation and prolonged hospital stays (lasting several weeks) [5][6][7]. Another basifying chemical which can be used is urease, which catalyzes urea (present in the urine) to carbon dioxide and ammonia, which will constantly increase the urine pH. This will lead to a slow dissolution of the uric acid stones. In instances of severe pain, urinary blockage, or other urgent scenarios, surgery becomes necessary to eliminate or disintegrate kidney stones, which is the more intensive of the methods mentioned above. While surgical intervention remains pivotal in treating kidney stones, its high recurrence rates and associated complications highlight the need for alternative minimally or non-invasive medical treatments focused on dissolving kidney stones and preventing their reappearance.

1.2 Miniature Robots for Medical Applications

Since medicine is moving toward minimally to non-invasive methods, miniature robots are becoming more and more popular, representing a fusion of advanced robotics and healthcare innovation. Miniature robots are small-scale robotic devices that are designed to perform specific tasks in confined or constrained environments. Their small size allows for precision and accessibility in delicate or complex operations, enabling them to navigate through intricate biological structures, reaching areas that were previously inaccessible or difficult to treat using conventional methods while being almost self-propelled, untethered, and controllable by an expert physician in clinical settings. They could have their propulsion mechanism on board, or more likely, they could be externally actuated.

One of the key advantages of miniature robots is their ability to deliver medications with high precision. By navigating through the body's complex networks, these robots can

be used for targeted drug delivery, localizing the administration of the drug and therefore reducing the side effects. The localized delivery of drugs increases the effectiveness of the treatment and reduces the dosage required, leading to improved patient outcomes and reduced healthcare costs.

In addition to drug delivery, miniature robots are designed to revolutionize minimally invasive surgical procedures. Traditional surgical techniques often involve large incisions and extensive tissue damage, leading to longer recovery times and increased risk of complications. Miniature robots on the other hand, can perform delicate procedures with remarkable precision, using small-scale tools and cameras to navigate and manipulate tissues with minimal trauma. These minimally invasive procedures help the patients to experience less pain, faster recovery, and lower chances of infection.

Furthermore, miniature robots show great potential in diagnostics and monitoring applications within bodies due to their potential biosensing function, meaning they have bioreceptors for the detection of various bioanalytes [8]. This continuous monitoring helps detect diseases earlier and improve the quality of treatment for the patients' personalized treatment plans, which leads to improved health outcomes and quality of life.

These robots can be manufactured using various biocompatible materials [9]. Furthermore, their size and shape can differ according to their application [10]. They can be made tethered or untethered, and having them untethered means that novel actuation methods are required to navigate them to their destination. These methods include the use of chemical reactions, optics, ultrasound waves, thermal reactions, magnetic fields, and many more. Having the advantage of the robots being untethered widens the spectrum for the usage of these robots as their range of movement grows. In this research, untethered robots were used, which maximizes their maneuverability as there is no catheter involved. Furthermore, this means that the robot can be held in place by just a magnetic patch with minimal likelihood of infections due to the existence of a foreign object inside the body.

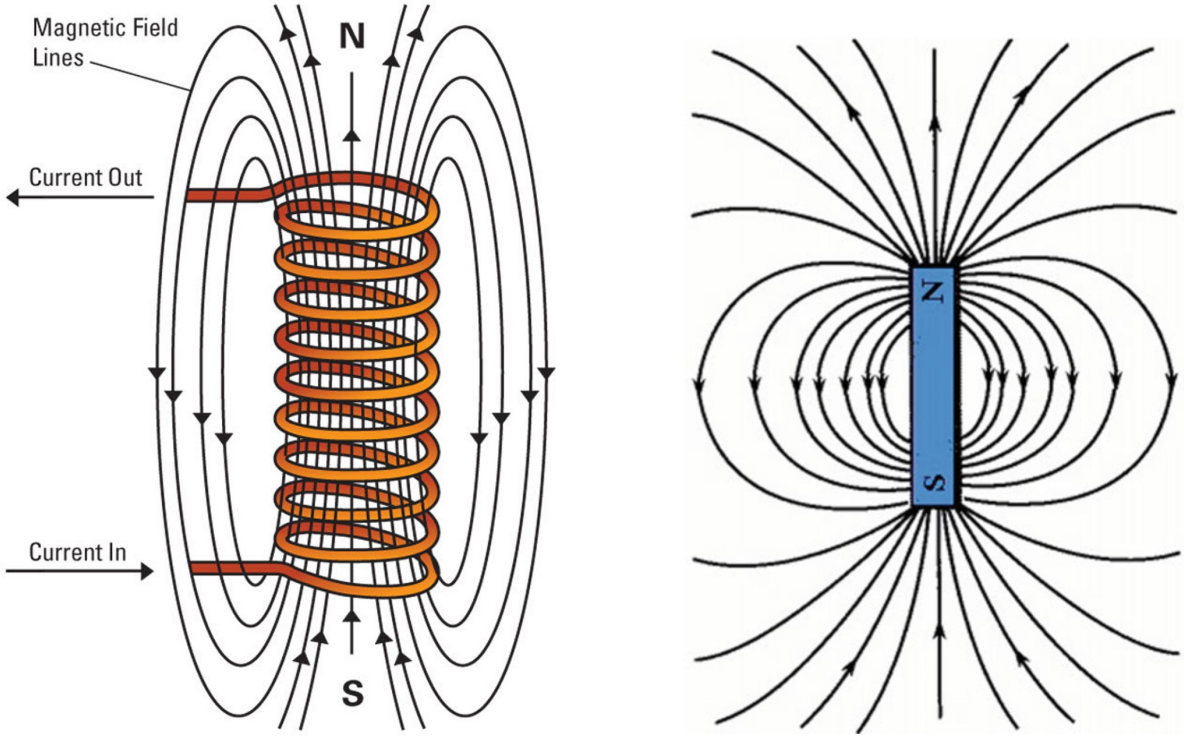


Figure 1.2: (a) Magnetic field lines produced by a coil [11] (© 2020 Springer Nature) (b) Magnetic field lines produced by a permanent magnet [12] (© 2018 Springer Nature)

1.3 Magnetic Actuation

Magnetic actuation is presented as one of the most used, reliable, and researched methods for untethered motion of microrobots. As stated in [13], static and slow time-varying fields, induce some effects on humans. Therefore, there are some limitations to the magnetic flux density that can be used. They also state that there are no lingering effects on the body in exposure to fields below 2T and if the individual is healthy, this number can increase

to 8T. It should be mentioned that these numbers differ in individuals with conductive or electrically active implants or pacemakers. In the case of dynamic fields with high frequencies, there is potential for the particles to be charged according to Faraday's law. Therefore, the field density should be monitored in these cases as well. In the review of the mechanisms represented in [13], it is stated that the magnetic fields normally used for actuation purposes are weaker than these limitations.

If the miniature robot is magnetized, an external magnetic field could generate enough force on it to create locomotion through attraction and repulsion with the magnets. Alternatively, a time-varying field such as a rotating field can also create locomotion, which in this case, is through the synchronization of the motion of the actuator and the miniature robot. As an example, if the external actuation magnet rotates, the robot also experiences this rotation and produces a rotary motion in the form of helical motion, drilling forward to swim, or in the form of a rolling ball on a surface. This external magnetic field, or the actuation magnetic field, can be produced by using electromagnets or permanent magnets. Having electromagnets means that the generated field can be very specific through the controlled generation of current, voltage, and field. However, the set-up of the electromagnet could be very complicated for a medical setting requires large workspaces as big as MRI machines, and create unwanted heat. MRI (magnetic resonance imaging) machines are widely used in medicine for imaging. They operate by inducing a magnetic field using a great coil with the appropriate frequency, exciting the atoms in the tissue and capturing the image by processing this excitation response received by another coil. On the other hand, permanent magnets could be used to create propulsion. In this case, there is some difficulty controlling the flux of the magnetic field as the magnetic circuit is not closed and there is flux leakage. However, this setup could be quite flexible as it only requires the presence and controlled movement of a permanent magnet. The strength of the field produced by the permanent magnet is significant enough to affect the miniature robots, and the workspace required is small.

1.4 Objectives and Contributions

The main objective of this thesis is to create a small-scale soft magnetic robot which supports the dissolution of kidney stones as well as prevents further stone formation. This main objective is broken down into the following three objectives.

1. Development of a magnetic actuation system based on permanent magnets, including the modeling, design, and fabrication of a magnetic actuation system capable of generating controllable rotating magnetic fields for guiding a miniature robot applicable for the clinical setting.
2. Design, fabrication, and characterization of a flexible miniature gelatin robot with optimized magnetization direction and location within the flexible filament.
3. Fabrication of drug-loaded gelatin robots and evaluation of their ability to increase urinary pH and support kidney stone dissolution (chemolysis) in vitro.

Achieving these objectives incurs the following contributions and potential impacts in the field of medical miniature robots and kidney health:

1. **Advancement of microrobotics for medical applications:**
   - Implementation of a magnetic control system with optimized magnet size for the best induction of force and torque on the robot, intended for clinical setting
   - Investigation of flexible filamentous miniature robots regarding their compatibility with applications in the urinary tract
   - Development of a design for the filament with the optimal magnet orientation and location
   - In vitro testing of the robot navigation inside anatomical organ models of the urinary tract
2. **Translational impact on health, especially in the urinary tract:**

- Targeted drug delivery in the urinary tract, reducing the side effects of systemic drug delivery such as gastrointestinal issues caused by orally administered medicine, or the side effects of antibiotics when administered systemically
- Dissolution of kidney stones using long-lasting potent enzymes loaded onto the filaments

1.5 Thesis Outline

The remaining chapters in this thesis are organized as follows:

Chapter 2 provides a comprehensive literature review, starting with an in-depth examination of magnetic actuation using electromagnets (2.1), covering fundamental design options and notable research. Section 2.2 describes magnetic actuation using permanent magnets, discussing their advantages, limitations, and specific use cases. Section 2.3 explores the design and fabrication of magnetic robots, focusing on design and locomotion techniques. This section highlights the latest advancements and challenges in the field. Finally, Section 2.4 discusses the applications of these robots in medicine.

Chapter 3 outlines the methodology used in this research. It begins with magnetic theory (3.1), describing the governing equations used for the propulsion of the miniature robots, their mechanical properties, and the hydrodynamics of the environment (3.2). Section 3.3 describes the overall actuation system, starting with an overview of the robotic arm (3.3.1). Section 3.3.2 discusses the rotating magnetic field mechanism, including its design, implementation, and role in achieving magnetic actuation. Section 3.3.3 details the camera and tracking system, used for monitoring and control of the robot's movements. Section 3.3.4 describes the urinary tract organ model used in the study. Section 3.4 discusses the robots' fabrication process and their application in kidney stone dissolution. Finally, Section 3.5 covers the experimental design, including locomotion and frequency response (3.5.1) and the dissolution of kidney stones (3.5.2).

Chapter 4 presents the experimental results and discussion. Section 4.1 focuses on the locomotion and frequency response of the robots, providing analysis and interpretation of the experimental data. Section 4.2 discusses the mechanical response of the filaments.

Section 4.3 highlights the efficacy of the robots in kidney stone dissolution under various conditions.

Chapter 5 concludes the research by summarizing the key findings and contributions of the thesis. It unifies the experimental results, theoretical insights, and practical applications discussed in the previous chapters. This chapter also suggests directions for future work, such as exploring advanced control algorithms, improving precision and efficiency, and expanding applications in other areas.

Chapter 2

Literature Review

As previously stated, the use of magnetic fields is a popular approach in the actuation of miniature medical robots. The magnetic forces and torques acting from the produced magnetic field of the actuator induce magnetized robot movement, envisioning application inside the human body thanks to the large penetration depth and safety of magnetic fields.

In this chapter, the state of the art for different methods of magnetic actuation, the structure of the robots, and their applications in the medical field will be discussed.

2.1 Magnetic Actuation: Electromagnets

One of the ways that an accurate magnetic field can be created, is the use of electromagnets. They can be designed to produce the exact magnetic field required for each application [14]. However, a few specialized designs exist which have proven to create a suitable field, force, and torque for actuation purposes.

One of these designs is called Helmholtz coil arrangement which consists of a pair of coils shown in Fig. 2.1.a. Helmholtz coils are known for creating an almost uniform magnetic field. In Helmholtz coils, two coils with the same radius are positioned at a distance of their radius apart, with their current in the same direction. Helmholtz coils were used in quite early research to actuate miniature robots. Sudo *et al.* used a two coil Helmholtz coil

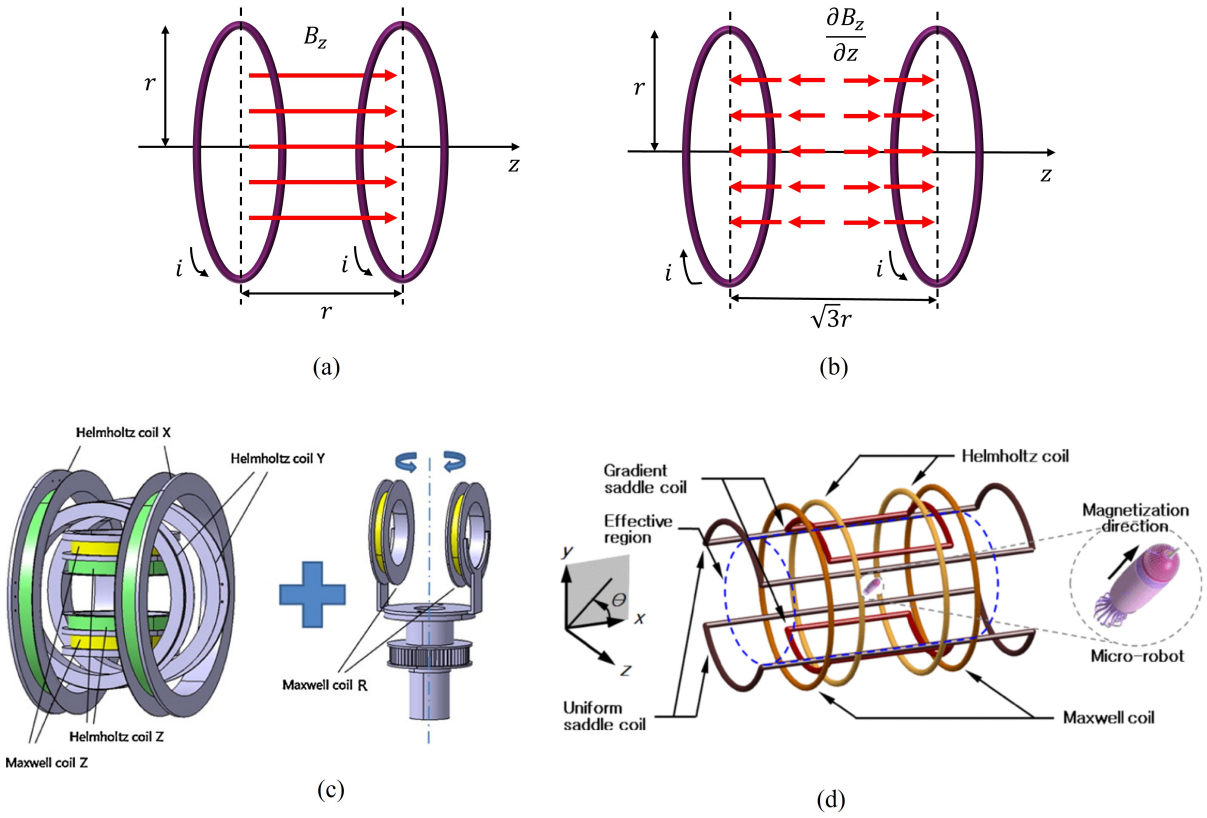


Figure 2.1: (a) Helmholtz coil pair (b) Maxwell coil pair (c) Set-up used in [15] consisting of 3 Helmholtz and 2 Maxwell pairs (Reprinted from the cited source © 2010, with permission from Elsevier) (d) Set-up used in [16] consisting of one Maxwell coil pair, one Helmholtz coil pair and one saddle coil (© 2010 IEEE)

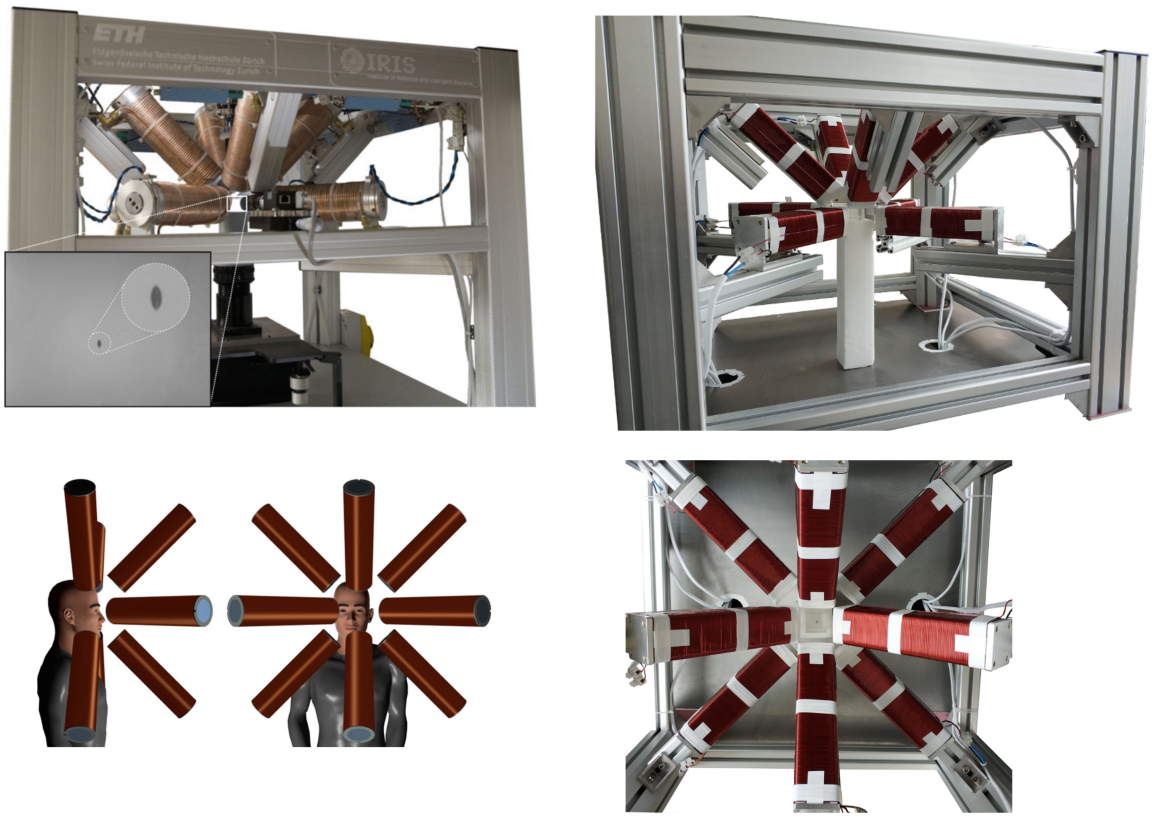
set up, creating an alternating magnetic field, to study the behavior of a miniature robot in viscous environments which showed promising results, achieving speeds as high as 80 mm/s and optimizing the shape of their robot [17]. However, the research quickly shifted to using more complicated coils for better control of the robots, one of them being Maxwell coils (Fig. 2.1.b). Maxwell coils have the same appearance as Helmholtz coils, with the difference being that the currents are in opposite directions and the distance between them is $\sqrt{3}r$. This creates a uniform magnetic field gradient, facilitating locomotion. Yu *et al.* introduced an electromagnetic-based actuation (EMA) system, which includes 3 stationary

pairs of Helmholtz coils, a stationary pair of Maxwell coils and a rotating pair of Maxwell coils (Fig. 2.1.c)[15]. The Helmholtz coils, with a uniform magnetic field, are used to align the robot to the required direction. The stationary Maxwell coils are used to produce propulsive force in the Z direction and the rotational Maxwell coils are used to propel the robot in the XY plane.

In another research, Jeon *et al.* proposes a system which utilizes a pair of Maxwell coils to generate a magnetic field gradient along the X axis, a pair of Helmholtz coils which create a uniform magnetic field along the X axis. The addition of the saddle coil adds a uniform field in the Y direction, meaning the robot is controllable in the XY plane without having rotation (Fig. 2.1.d) [16].

To achieve an even higher level of control, multiple coils could be arranged, mathematically modeled and activated to induce the exact amount of field or gradient needed to align or propel the robot in any direction. OctoMag is a sophisticated electromagnet configuration designed for precise control of magnetic microrobots in a 5-degree-of-freedom (DOF) manner (Fig. 2.2.a) [18]. Consisting of four electromagnets in an upper set and four in a lower set, OctoMag enables vertical lifting and horizontal motion control. Each coil within this system is strategically positioned to generate specific forces and torques, allowing for versatile manipulation of the microrobot's orientation and position. By optimizing coil arrangements and current flow, OctoMag achieves fine-tuned control, essential for applications requiring intricate movements in constrained spaces. Unlike the systems mentioned before, Octomag is not reliant on a uniform field or gradient for actuation, instead, the positioning and alignment of the robot is achieved through a fine-tuned control system. As a way to simplify the control system of Octomag, Yuan *et al.* suggest a set-up of 8 rectangular coils (Rectamag3D), able to control the robot the same way, with the advantage being that rectangular coil set-ups have exact mathematical models, therefore, the forces and torques induced by them are simpler to calculate, facilitating a closed-loop control system to control and propel the robot [19] (Fig. 2.2.b). It is mentioned that the use of coils is safer in a clinical setting since the field can be cut off completely, however, depending on the application and the clinic, that might not be the case.

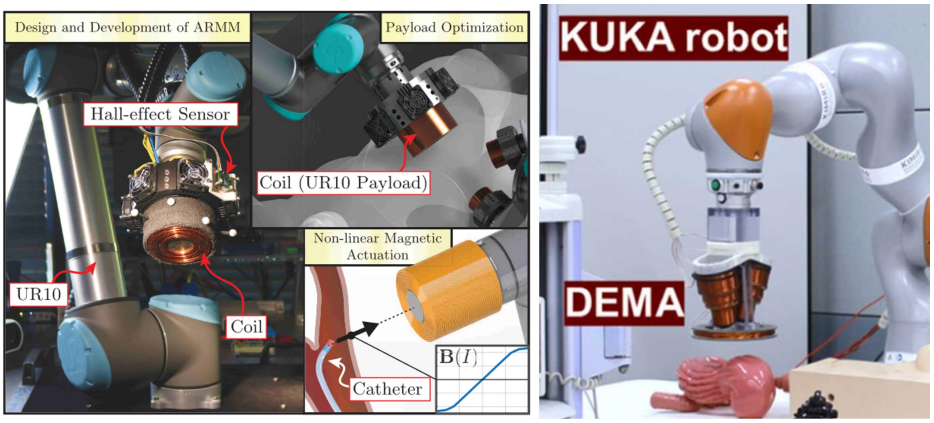
In all the set-ups explained before, the actuation system is almost completely stationary, this means that the workspace of the actuation system is a designated area in the middle



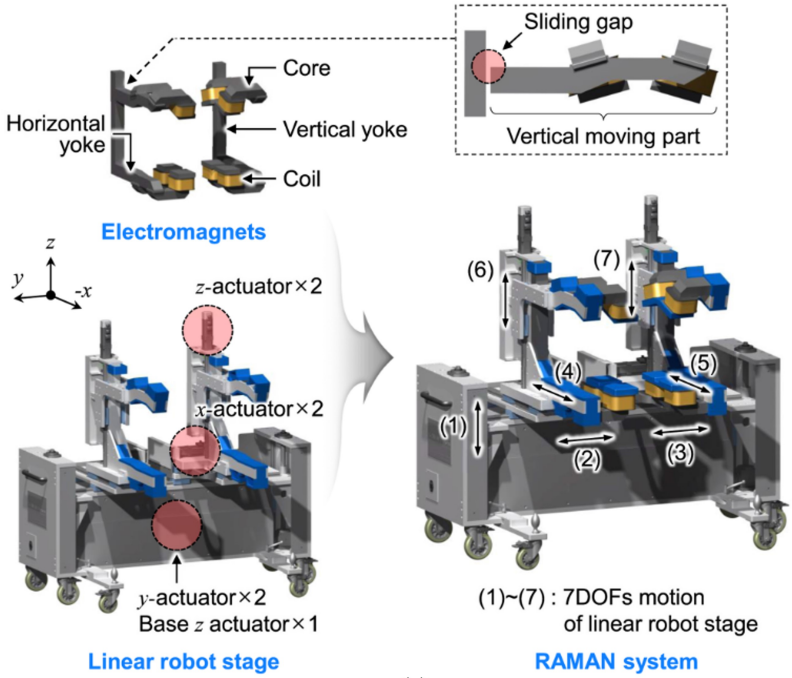
(a) (b)

Figure 2.2: (a) Octomag [18] (© 2010 IEEE) (b) Rectamag3D [19] (© 2020 IEEE)

of the coils. This set up is less than optimal for biomedical applications, since coils as big as MRI machines are needed to have a precise workspace in the human body. Additionally, it is difficult to achieve sufficient strength to produce locomotion in that workspace. In clinical applications of magnetic actuation, the workspace needs to be variable, which is difficult to achieve using coils. This means it is preferred to have the robot and actuator moving simultaneously and the actuator guiding the robot to the desired location in the human body. The downside in this scenario is that there would be a great amount of leakage flux, making the system not as efficient, however, the field would still be strong enough to actuate the robot.



(a) (b)



(c)

Figure 2.3: (a) ARMM [20] (© 2023 IEEE) (b) DEMA [21] (Reprinted from the cited source © 2023, with permission from Elsevier) (c) RAMAN [22] (© 2024 IEEE)

Research in recent years suggests the electromagnet to be mobilized by some type of robotic arm. Riccardi *et al.* introduced a groundbreaking magnetic actuation system named the Advanced Robot for Magnetic Manipulation (ARMM) system (Fig. 2.3.a) [20]. Unlike traditional set-ups, ARMM employs a single, mobile electromagnetic coil integrated into a 6-degree-of-freedom robotic arm, offering an expansive workspace due to its coil mobility. This system overcomes limitations posed by stationary electromagnets, providing a wider range of magnetic fields which are controllable and have the ability to be deactivated if necessary. The design process of ARMM includes optimizing the electromagnet to generate maximum magnetic fields within a defined heat dissipation constraint, crucial for safe and effective operation. To tackle nonlinearities in the coil's behavior at high currents, a sophisticated strategy utilizing Hall effect sensors for real-time correction and control was proposed, ensuring precise magnetic actuation even beyond the coil's saturation limits. A Hall effect sensor is a device that measures magnetic field strength, where a voltage is generated perpendicular to both the current flow and the magnetic field. This voltage is proportional to the strength of the magnetic field, allowing the sensor to detect and measure it accurately. Moreover, ARMM enables codirectional control of magnetic fields and forces, independently adjusting their magnitudes. This capability is instrumental for steering flexible surgical instruments, such as endovascular catheters, over extensive distances with pinpoint accuracy. The system's integration with a collaborative robotic arm like the UR10 further enhances its adaptability and safety in clinical environments.

In another case, Hoang *et al.* introduces a novel Dual Electromagnet Actuator (DEMA) integrated with an articulated robotic arm for controlling a Magnetic Capsule Endoscope (MCE) with 5 degrees of freedom (Fig. 2.3.b) [21]. The system employs magnetic induction for localization using transmitter and receiver units, along with a control methodology which is user-friendly and is based on position feedback and images from the camera. Testing includes 3D closed-loop and open-loop control in in-vitro and stomach phantom experiments which are then validated by in-vivo studies on pig cadaver stomach and fresh small intestine segments, showcasing multi-functionality with active biopsy capability. The DEMA design involves optimizing electromagnet structure, actuation method, and dynamic modeling with stability analysis, leveraging MATLAB's fmincon for optimization and COMSOL Multiphysics for Finite Element Method (FEM) simulations. Actuation is

achieved in 5-DOF, with control in a plane through the magnetic field and robotic arm rotation, ensuring safety, low power consumption, and active locomotion while maintaining an open configuration and multi-functionality for diagnostic purposes. The actuation mechanism's mathematical modeling enables precise control of the MCE's motion, offering potential advancements in endoscopic procedures and clinical applications.

Most recently Lee *et al.* argues that all the developments mentioned lack the controllability of either the size and position of the workspace or the strength of the magnetic field, therefore they introduce a system called RAMAN [22]. The RAMAN system, shown in Fig. 2.3.c addresses this gap by introducing a novel structure consisting of two electromagnets and a linear robot stage. Each electromagnet features a sophisticated design comprising vertical yokes, horizontal yokes, and cores and coils, all made from soft-magnetic materials to ensure a closed-magnetic circuit and optimize magnetic performance. The linear robot stage, equipped with seven actuators, enables precise control over the position and size of the workspace through coordinated motion in multiple axes. This comprehensive approach not only enhances magnetic performance but also ensures compatibility with medical imaging devices like C-arm X-ray systems, crucial for real-time visualization during medical interventions. The RAMAN system's effectiveness is validated through a prototype that demonstrates compact size, manageable weight, and compatibility with existing medical infrastructure, making it a promising solution for advanced medical magnetic robotics.

2.2 Magnetic Actuation: Permanent Magnets

An alternative to using electromagnets in magnetic actuation is the use of permanent magnets. In this case, the field is not as customizable as electromagnets, and there is flux leakage from the magnet, meaning as the magnetic circuit is not closed, the efficiency is not very high. However, through optimization, a suitable permanent magnet can be selected for each application. Permanent magnets are able to generate a field which is 10 to 20 times higher and a gradient 2 to 3 times higher than an electromagnet, if they are the same size and the current is kept at a safe number [23]. The field of an electromagnet is dependent on its current and the size of the coil, whereas the main factor in a permanent magnet

is its magnetization vector. As an example, imagining we have a permanent magnet and an electromagnet of the same size with the same field strength, and then cut them in half, the field strength at the surface of the permanent magnet stays the same because it is dependent on the magnetization of the material. However, for the electromagnet, the field is dependent on the number of turns for the coil and we would have half the number, so the field would have half the strength of the original coil. The results obtained by Erni *et al.* show that same-size permanent magnets show more promise for clinical applications due to the limitations of the feeding current in an electromagnet. Furthermore, a permanent magnet is simpler than electromagnets and therefore, is easier to transport to a clinical setting, and set up the workspace for the human body in a patient-friendly environment. Lastly, a clinical setting needs a controlled temperature and an environment whereas electrically charged coils produce heat.

The permanent magnet, same as the electromagnet, can either produce a constant and unchanging magnetic field, in which the robot moves through attraction and repulsion with the actuator, in which case the magnet drags its target and for that to happen, the magnetic field needs to be particularly strong and very close to the miniature robot. Alternatively, permanent magnets can produce a constant magnetic field gradient through rotation, meaning the direction of the field varies with time. This field induces such torque or force on the target so that it tries to sync with the actuator, creating a propulsive force. In this case, the motion of the robot is its propulsion mechanism which helps its locomotion. Therefore, it can be stated that the robot is more self-propelled. In this case, the magnetic field does not need to be as strong or as close to the miniature robot, creating more flexibility in the workspace.

The most common format of using a permanent magnet is attaching the magnet to a robot arm, which means its position and orientation is programmable and controllable at all times. Because of the simplicity of the set-up, the research including a permanent magnet mostly focus on the design of the miniature robot and the application and control methods.

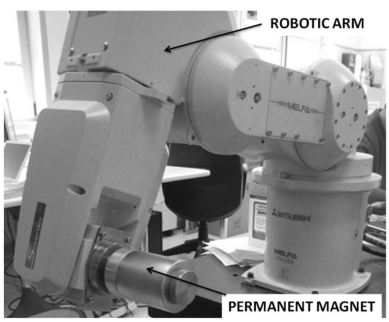
In early works Ciuti *et al.* propose using the combination of an external magnet, and an internal magnet placed inside an endoscopic capsule for investigations in the GI tract. In this case, the actuator magnet is stationary, and only magnetic forces are used to pull

Feature	Electromagnets	Permanent Magnets
Magnetic Field	Customizable	Fixed
Field Strength	Lower (10-20x)	Higher
Field Gradient	Lower (2-3x)	Higher
Power Consumption	Requires continuous power	No power consumption
Setup Complexity	More complex	Simpler
Controllability	Highly controllable	Limited controllability
Heat Generation	Generates heat	No heat generation
Workspace	Limited by coil size	More flexible workspace
Cost	Can be expensive	Generally lower cost

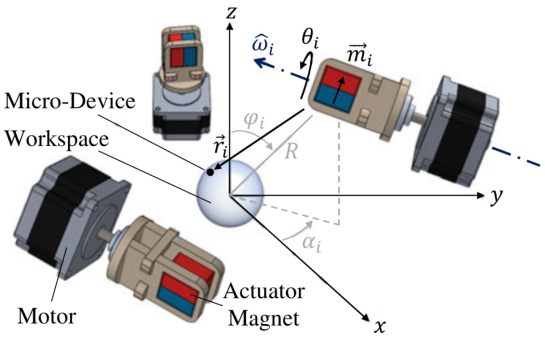
Table 2.1: Comparison of Electromagnets and Permanent Magnets

and push the endoscope [24]. A user-friendly Human Machine Interface (HMI) allows the medical professional to control the system while a robotic arm positions the external permanent magnet, which interacts with the internal magnets. Inertial sensing feedback is incorporated to enable control and positioning accuracy. In other work, Mahoney and Abbott present a method for controlling the rotation of untethered magnetic devices using a single rotating permanent magnet [25]. The state of the art in this research demonstrates that the desired rotation of the magnetic field at the robot's position can be achieved from any location of the actuator magnet, as long as its rotation axis is adjusted accordingly, which is then verified through experiments. The research team further developed a mathematical model and control algorithm to maneuver an endoscope via magnetic forces using a permanent magnet on a robotic arm, achieving 5 DoF for the movement of the endoscope [26].

An alternative to having only one rotating permanent magnet is to have a set-up of multiple rotating permanent magnets, which Ryan and Diller state is less hazardous than having multiple moving parts and has the potential to strengthen the magnetic field which controls the miniature robot [27]. In this case, the authors prove that having 6 permanent magnets in the actuation set up is necessary for a singularity-free control where any combination of forces and fields is achievable to have a 5 DoF control over the miniature robot.



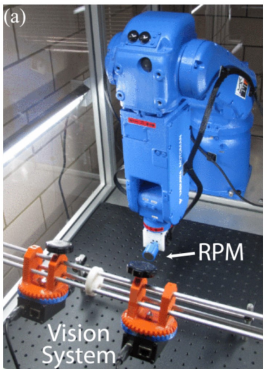
(a)



(b)



(c)



(d)

Figure 2.4: (a) Single magnet to push and pull [24] (b) Multiple rotating magnets [27] (© 2017 IEEE) (c) The Stereotaxis Niobe Robotic Magnetic Navigation System [28] (© 2007 John Wiley and Sons) (d) Single rotating magnet [25] (© 2014 IEEE)

Their prototype takes the set-up further to having 8 permanent magnets, optimized and controlled for enhanced performance and reduced motor demands in challenging scenarios. The use of more than one permanent magnet is also proposed in [29] [30] [31], stating that the advantage could be stronger fields and, depending on the configuration, more control over torque or force. Most importantly, Armacost *et al.* introduced a magnetic actuation system consisting of two big permanent magnets which are commercially used to insert

catheters used in cardiology [28].

The focus of recent research has mostly been on developing a control strategy for actuation using permanent magnets. Researchers very often mention that modeling a permanent magnet is difficult, therefore, it is difficult to have a control algorithm for them, mentioning their different approaches. However, permanent magnets are attractive in set-ups thanks to the simplicity of implementation. The authors in [32], [33], [34], [35] and [36] discuss various different control and localization strategies to implement permanent magnets for applications in physiological environments.

In summary, a comparison of electromagnets and permanent magnets is included in Table 2.1, regarding their benefits and drawbacks for same-size magnets keeping in mind the restrictions of current fed to the electromagnets.

2.3 Magnetic robot design and fabrication

The other side of the actuation is the design of the robot, which is based heavily on its application. There have been no miniature robots for applications in the urinary tract. However, there has been extensive study done on various sizes of miniature robots, mostly in the micro-scale, for different medical applications. These applications include capsules and catheters used in the GI tract, screw-like and abrasive robots to target blood clots, robots for applications in a lumen and many more. The mechanisms observed in these robots are mostly pull and push motion, helical rotation and rolling motion, however, more complex movements could be designed and achieved as well.

The most often discussed mechanism in the literature is helical rotation, similar to a motion produced by flagella in microorganisms, where the external actuator magnet is placed in parallel with the robot's magnet, and the field's rotation causes a helical motion on the robot, which is the driving mechanism similar to bacterial flagella. Park and Park introduce a microrobot designed with a cylindrical body and equipped with helical structures which enable its propulsion [37]. This helical propulsion mechanism is particularly effective for navigating through viscous and confined environments like the human vascular system. Similarly, Fountain *et al.* developed a rod-shaped device with this rotational

motion [38]. Recently Khalil *et al.* introduced a fully applicable system which includes a helical screw-shaped robot, 1 mm in diameter, controlled by a permanent magnet-based robotic configuration alongside an array of Hall-effect sensors [33]. This helical robot, incorporating a cylindrical permanent magnet, engages with rotating dipole fields produced by synchronized DC motors linked to permanent magnets. This interaction induces a regulated magnetic torque, propelling the helical robot within fluid-filled environments like catheter segments or rabbit aortas. Below these environments, strategically positioned Hall-effect sensors detect magnetic field variations stemming from the helical robot and the rotating dipole fields. These sensor readings, combined with a pre-computed magnetic field map, are leveraged to gauge the helical robot's position relative to the sensors. Subsequently, closed-loop motion control is used, combining the estimated position with control algorithms to synchronize the rotating dipole fields, modulate the robot's motion, and facilitate precise navigation and interaction with targets such as blood clots. Experimental verification in in-vitro and ex-vivo settings underscores the system's effectiveness in localization, motion control, and target engagement.

Another form mimicking bacterial flagella was reproduced by Sudo *et al.* Their magnetic swimming mechanism consists of a permanent ring magnet and a film tail with a thickness of 0.1 mm [17]. The operational principle involves magnetic torque acting on the permanent magnet when exposed to an external magnetic field, where the magnetic field varies sinusoidally in magnitude. This alternating magnetic field causes the permanent magnet to exhibit rotational oscillation, which in turn induces bending motions in the robot's tail. These bending motions propel the robot forward by pressing the tail against the surrounding fluid (Fig. 2.5.a).

Achieving free and efficient motion of microrobots on unknown surfaces is challenging, and there is a close alternative to the helical rotation motion which overcomes this difficulty. Hou *et al.* introduce a rolling locomotion method for a magnetic microrobot with a rectangular body, measuring approximately $300\,\mu\text{m} \times 200\,\mu\text{m} \times 50\,\mu\text{m}$ [39]. This method relies on an external rotating magnetic field, leveraging magnetic, normal, and friction forces to induce consecutive rotations on the microrobot. A magnetic field rotating at 2 rps propels the microrobot at 1.4 mm/s, allowing it to move along lines or curves and climb slopes or stairs. Jiang *et al.* also propose a rolling microrobot which utilizes an external magnetic

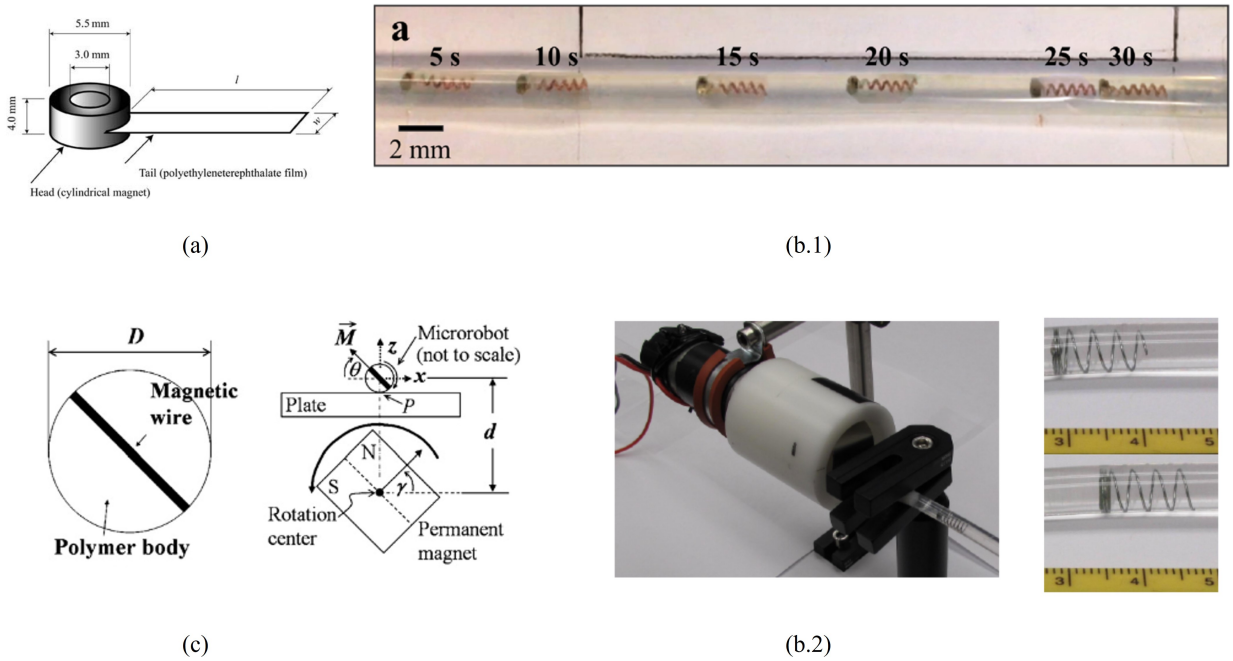


Figure 2.5: (a) Oscillatory fin on a microrobot [17] (© 2006 Sage Publications) (b) Helical motion in a tube 1.[33] 2.[38] (© 2010 IEEE) (c) Rolling motion [39] (© 2010 AIP Publishing)

actuator in the form of a rotating permanent magnet with a rectangular body [40]. The magnetization of the microrobot occurs along its long axis due to the hysteresis of the ferromagnetic wire. The rotation of the magnet leads to the rotation of the microrobot, causing it to maintain the magnetization aligned to the external field direction. Their robot is fabricated by coating an iron wire with UV adhesive to form micro balls, which are then hardened and separated to create individual microrobots. These microrobots have diameters around 400–440 μm and encapsulate an artificial easy axis for controlled rolling motion. Similarly, Mahoney and Abbott designed a robot which moves through the same mechanism with its body being ball-shaped [41] (Fig. 2.5.c).

Another way to look into the design of the microrobots is the ones which take the general shape of a rod, where the placement of the magnet and the material properties,

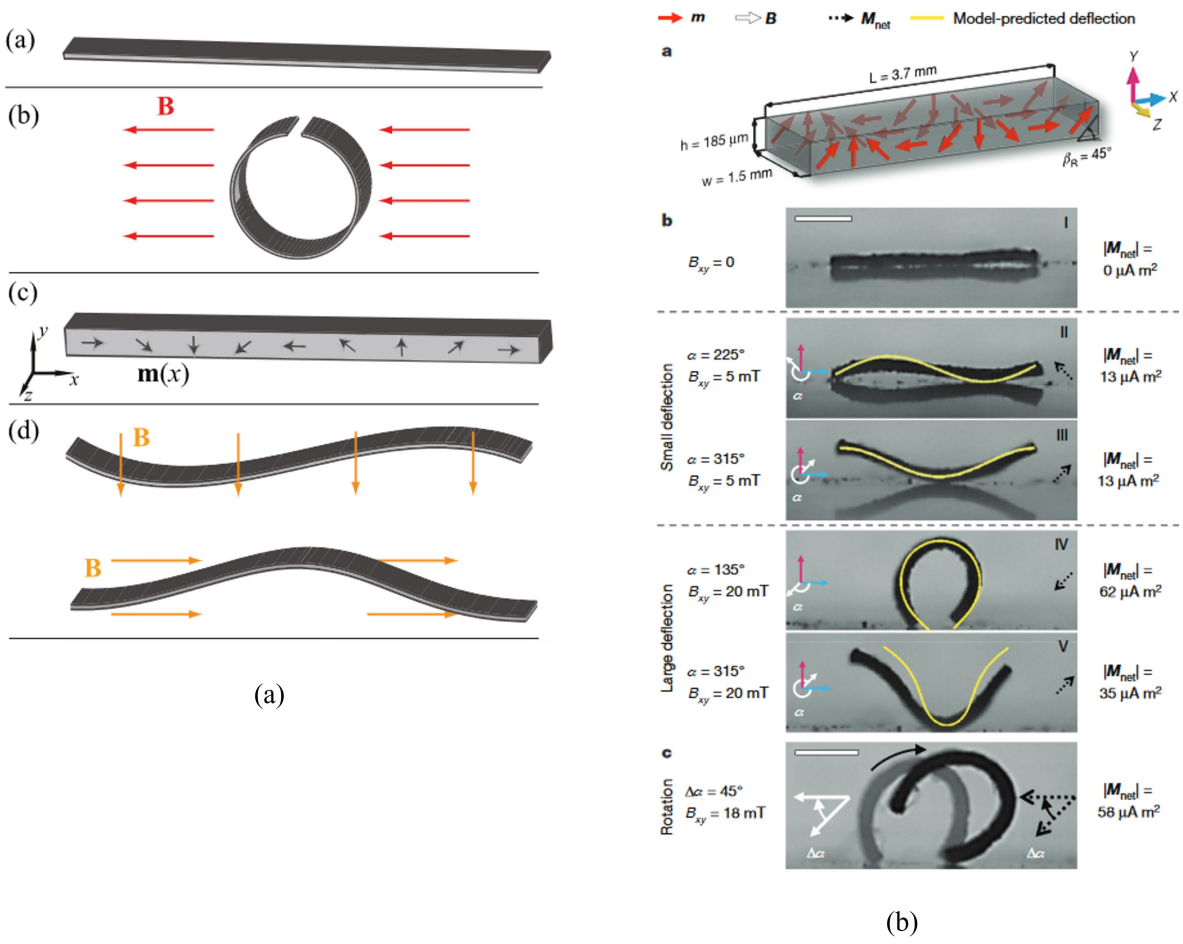
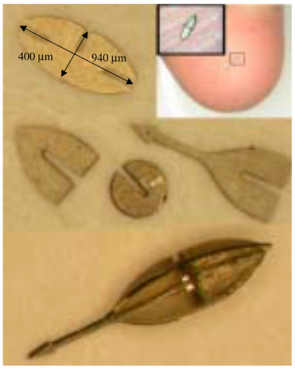


Figure 2.6: The rod-shaped magnetic robot design of Diller *et al.* (a) [42] (© 2014 AIP Publishing) and Hu *et al.* (b) [43] (© 2018 Springer Nature), and their deformation in different magnetic fields.

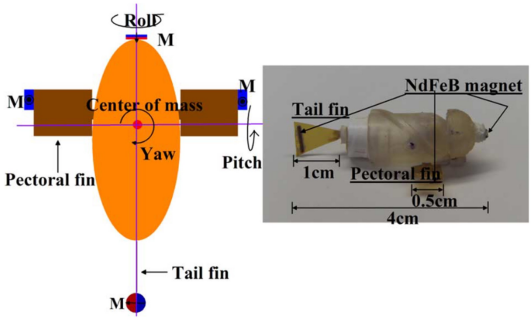
could produce more complex motions. Diller *et al.* introduce a millimeter-scale robot capable of generating a traveling-wave deformation along its body using a spatially varying magnetization profile and an external rotating magnetic field [42]. The robot's construction involves a flexible sheet made of neodymium-iron-boron (NdFeB) magnets mixed with platinum-cure silicon rubber (Ecoflex), enabling it to deform in a sinusoidal wave pattern when subjected to the magnetic field. The magnetization profile, wavelength, and

field frequency significantly influence the robot's swimming speed, with Taylor's model providing an initial approximation of this relationship [44]. Experimental results demonstrate a linear correlation between field strength and deformation amplitude until larger deformations occur. Evaluation in both water surface and underwater conditions reveals varying swimming speeds based on field parameters, with discrepancies attributed to fluid immersion and wall effects not fully accounted for in the model. This development paves the way for micrometer-scale soft robots. A similar design is by Hu *et al.* [43]. The soft robot described in their work is constructed from a blend of silicone elastomer and NdFeB microparticles, giving it magneto-elastic properties that enable shape changes and locomotion in response to external magnetic fields. The robot's design includes a single-wavelength harmonic magnetization profile along its body, allowing it to be controlled by a time-varying magnetic field to produce different modes of locomotion. The magnetic field interacts with the robot's magnetized body, producing spatially varying magnetic torques that change the shape of the robot and enable steering in desired directions. For instance, small-magnitude magnetic fields aligned along principal directions create sine or cosine shapes, while high-magnitude fields aligned differently cause the robot to deform into 'C' or 'V' shapes. Rotating the magnetic field induces rigid-body rotations, enabling modalities like rolling, walking, and jumping. These mechanisms are harnessed for tasks such as swimming, meniscus climbing, crawling, and even navigating complex terrains, showcasing the versatility and potential applications of such soft robots in controlled environments.

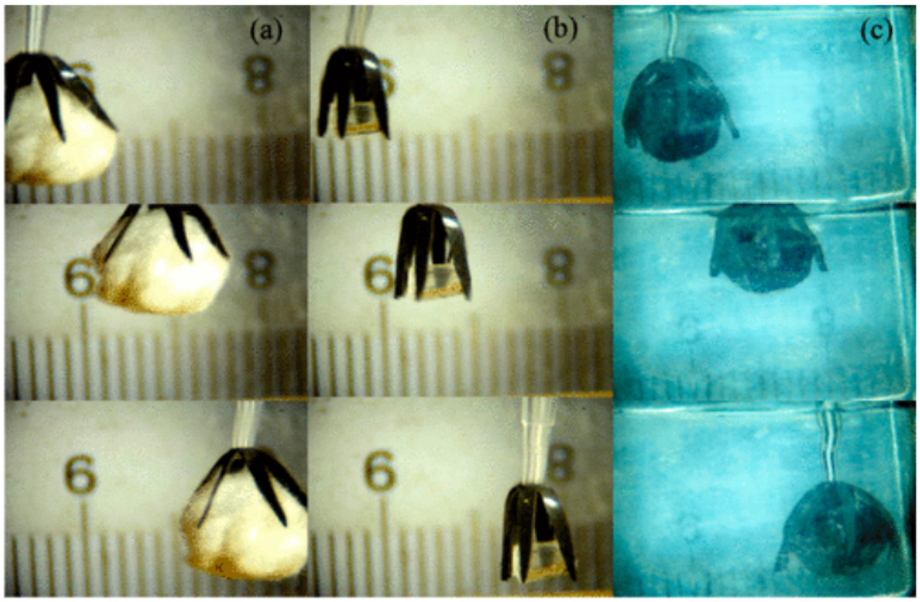
One of the first non-helical, complex designs was introduced by Yesin *et al.* [45]. The robot is constructed from a combination of four basic 50 µm thick nickel parts bonded with UV-activated glue, resulting in a winged-ellipsoid shape with an axis of symmetry along its long axis. This design allows an external magnetic field to effectively direct and move the robot along its axis due to shape anisotropy, similar to the way a needle aligns along its long axis when magnetized. Additionally, the winged shape minimizes sideways drift by increasing fluid drag along the perpendicular axes. The microrobot's general dimensions include a thickness of 50 µm for the nickel parts. This design introduces different modalities of motion in the body of the robot. Later Kim *et al.* design another fish-robot inspired from biology which is propelled by a single flexible magnetic actuator employing a rotating magnetic field in a three-axis Helmholtz coil set-up [46]. Typically, magnetic



(a)



(b)



(c)

Figure 2.7: (a) Ellipsoid magnetic microrobot [45] (© 2006 Sage Publications) (b) Fish-like magnetic microrobot [46] (© 2012 IOP Publishing) (c) Flower-like soft microrobot [47] (Reprinted from the cited source © 2016, American Chemical Society)

fish-robots utilize alternating and gradient magnetic fields, enabling limited motion such as fin bending. However, the use of a flexible magnetic actuator which is driven by an external rotating magnetic field allows for multiple gaits, including bending vibration, twisting vibration, and their combinations. Unlike conventional magnetic fish-like micro-robots lacking pectoral fins, this robot incorporates pectoral fins and magnetic torque direction control to achieve swimming and maneuvering capabilities. Their research emphasizes the correlation between the robot's dynamic actuation, magnetic actuator, and rotating magnetic field. Key features include 6 degrees of freedom, a novel control method ensuring posture stability, three-dimensional swimming, improved velocity control, and enhanced turning abilities. They also discuss the driving method and multi-motion control of fins using a rotating magnetic field, leading to diverse oscillatory motions and turning performances. The proposed magnetic fish-robot demonstrates promising advancements in biomimetic locomotion and serves as a potential model for diagnostic robots, showcasing the integration of innovative materials and control strategies for enhanced micro-robotic capabilities.

A very interesting mechanism introduced in research is by Gao *et al.*, the design of the flowerlike soft platform which revolves around a silicon rubber matrix infused with ferromagnetic particles, providing a responsive and flexible structure [47]. This platform is fabricated using a combination of 3D laser printing and composite material techniques, resulting in a lightweight and adaptable structure reminiscent of flower petals. The geometry of the platform includes variations in size, length, and thickness of the petals, allowing for different levels of deflection and response to external magnetic fields. The locomotion function of this platform is driven by a magnetic force acting on the petal, which causes a deflection in the presence of an external magnetic field gradient. This deflection mimics the natural movement of flower petals responding to light or other stimuli. Importantly, the platform's intrinsic elasticity, coupled with the magnetic force, enables it to return to its original shape when the magnetic field is removed, showcasing a reversible and controllable locomotion function. This functionality allows the platform to grasp and transport objects of varying shapes and sizes in air or liquid environments, demonstrating its versatility and potential for applications in micro/nanorobotics, biomimetic research, and drug delivery systems.

Overall, the literature shows that there are ways to make the designs complicated, however, basic rolling and rotating designs are still prevalent with them moving towards the use of soft materials for ease of movement and actuation. This also makes carrying drug loads possible for the robot through absorption and release, which happens in a hydrogel like a sponge.

2.4 Medical Applications of Magnetic Robots

The previous section was a comprehensive discussion of the different geometries and motion mechanisms of magnetic miniature robots. However, their application is also a significant topic. As mentioned in the introduction, magnetic miniature robots have great potential to be used in the biomedical field due to the compatibility of the magnetic field required with the human body. They could be used as swimmers and crawlers in the gastrointestinal (GI) tract, brain, spinal cord, blood capillaries, inside the eye and the urinary tract to provide minimally invasive care for the patients. Sitti *et al.* has clearly categorized the applications of these robots, which will be discussed in this section in the same manner [10].

Visual imaging techniques like endoscopy and laparoscopy are critical for disease diagnosis. Traditional flexible endoscopes and catheters, while effective, are invasive and only suitable for short-term use. Capsule endoscopes, which are pill-sized and less invasive, provide a significant alternative, especially for long-term imaging and accessing previously unreachable areas like the small intestines [52]. These devices, equipped with cameras and wireless transmission, currently offer passive imaging. Transforming them into active capsule robots would enhance their capabilities, allowing for controlled, untethered imaging of difficult-to-reach areas over extended periods. Localization of these capsules within the GI tract, is useful for precise diagnosis and advanced techniques like 3D visual mapping. Methods include medical imaging devices like fluoroscopy, ultrasound, PET, and MRI [53]. A notable example is the magnetically actuated soft capsule endoscope (MASCE) with a CMOS camera, designed for safe, controlled imaging inside the stomach [24].

Currently, passive biomedical sensors implanted in or on the human body monitor

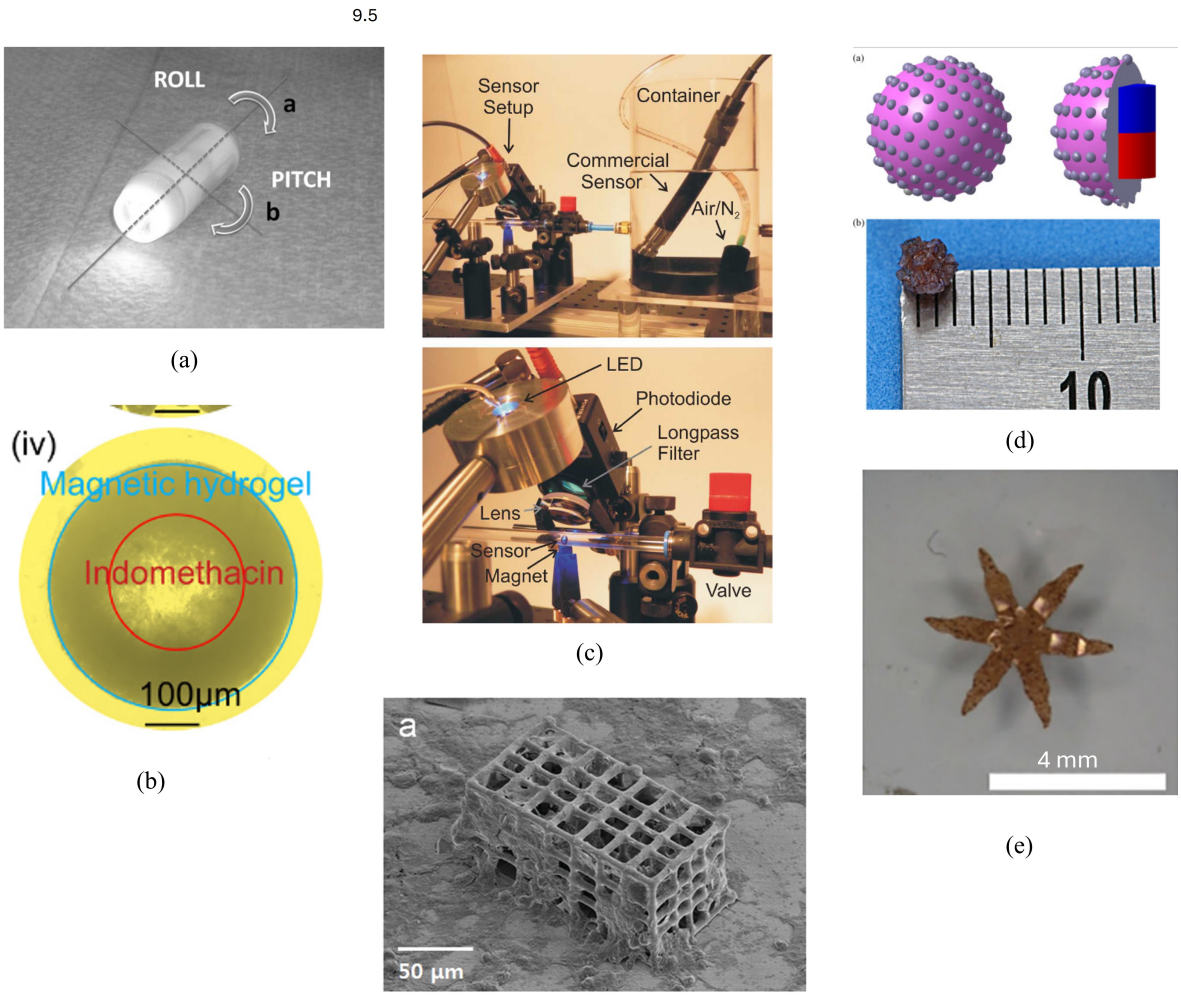


Figure 2.8: (a) Endoscopic capsule [24] (b) Hygdogel spherical robot for intravascular drug delivery [48] (© 2023 Royal Society of Chemistry) (c) Magnetic optical oxygen sensor [49] (© 2008 IEEE) (d) Rough spherical magnetic miniature robot for drilling and removing blood clots [15] (Reprinted from the cited source © 2010, with permission from Elsevier) (e) Soft gripper robot [50] (© 2017 IEEE) (f) Magnetic microrobot for stem cell culture and delivery [51] (© 2013 John Wiley and Sons)

health conditions by detecting various parameters like glucose, pH, temperature, and neural activity. Integrating mobility and control into these sensors using milli/microrobots could enable a mobile medical sensor network for active health monitoring, patrolling different body parts in a minimally invasive manner. Although this application is still underexplored, initial studies have demonstrated its potential, such as a magnetically controlled microrobot for optical oxygen sensing inside the eye [49].

Targeted therapy aims to deliver therapeutics like drugs, genes, or stem cells to specific body regions, minimizing side effects and enhancing treatment efficacy. Miniature robots can release these substances precisely and in controlled amounts. Active capsule endoscopes have been used for targeted drug delivery in the GI tract, using various triggering mechanisms for controlled release [54] [24]. Studies have also explored using miniature robots for drug delivery in blood vessels, employing magnetic fields for navigation and control [48].

Milli/microrobots also hold potential for minimally invasive surgeries, such as tissue removal. Additionally, robots have been used for tasks like puncturing blood vessels and potentially opening clogs in blood vessels [33] [15]. These applications could significantly expand with further research, offering new surgical possibilities within the circulatory system, brain, and other organs. Other types of procedures could be done by grippers, such as the one mentioned in [50].

Miniature robots can precisely deliver stem cells and regenerate tissues at pathological sites, aiding in treating various diseases. Initial research has developed the robots for this purpose, such as a cage-shaped microrobot for cell culture and tissue regeneration [51].

Chapter 3

Methodology

This work aims to introduce soft filamentous miniature robots which are loaded with appropriate drugs for applications in the urinary tract. The robots are coupled with an external actuation system, which will also be discussed. An overview of the system is shown in Fig. 3.1.

As previously discussed, the use of coil setups to generate a magnetic field for actuation is a viable option. However, despite their simplicity and controllability, achieving precise alignment of the workspace, especially within the confinements of patient organs, is a significant challenge. Consequently, in such scenarios, the use of a rotating permanent magnet is presented as a more favorable alternative, which is modeled, and built for this work. The proposed configuration involves mounting a permanent magnet onto a DC motor whose position is controlled by a robotic arm, facilitating the generation of the magnetic field for actuating the filamentous miniature robots. This set-up not only ensures precise manipulation within the anatomical complexities of the urinary tract but also highlights the system's adaptability to dynamic clinical settings.

The other components of the system, i.e. the filamentous robots, are gelatin-based 12 mm long, 1mm thin filaments in which a permanent micromagnet is placed. The dimensions of the robot are specifically tuned to the urinary tract cavities. The interaction of this permanent magnet, referred to as the robot magnet, with the actuator magnet,

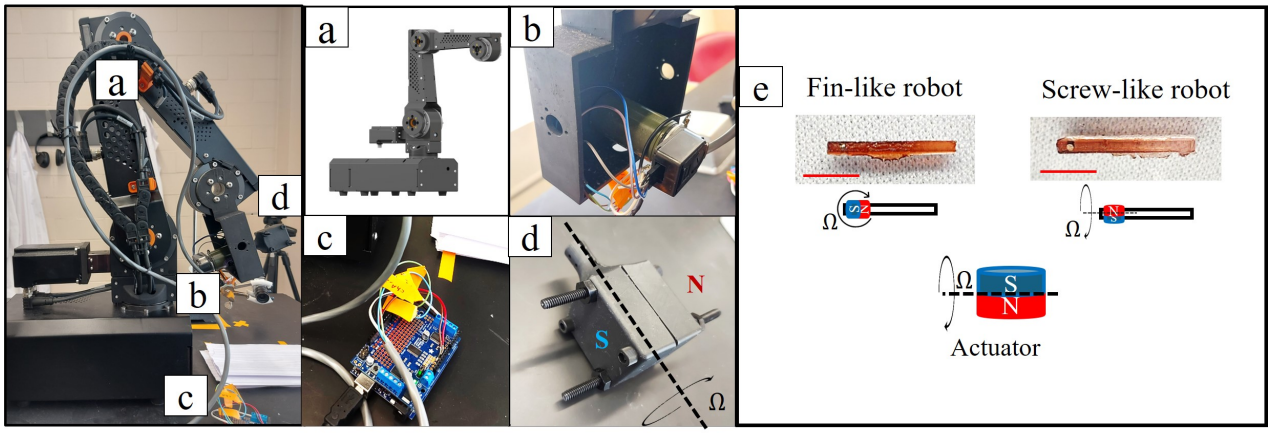


Figure 3.1: (a) Robot arm from igus (b) DC motor attached to the end-effector of robot arm (c) Arduino Uno as microcontroller to monitor the RPM (d) Permanent magnet inside a holder rotated by the DC motor (e) Proposed configurations for the soft filament miniature robots

creates propulsion for the robots. In this study, the effects of the magnet location and orientation in the filament on the performance of the robot will also be studied.

3.1 Magnetic Theory

Using magnetic actuation in robotics means using the forces and torques which are induced on a permanent magnet in a magnetic field to create its locomotion, making untethered locomotion of small-scale robots possible. A magnetic field surrounding a permanent magnet creates the following force and torques

$$\vec{F} = V_m(\vec{M} \cdot \nabla)\vec{B} \tag{1}$$

$$\vec{T} = V_m(\vec{M} \times \vec{B}) \tag{2}$$

in which V_m is the volume and $\vec{M}$ is the magnetization vector of the permanent magnet. Equation (1) describes the force ($\vec{F}$) exerted on the magnet due to the magnetic field

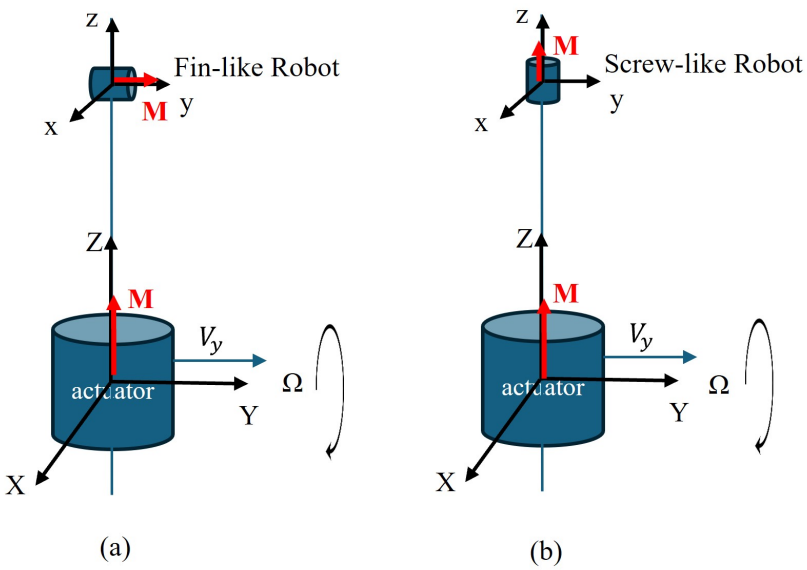


Figure 3.2: (a) The actuator magnet and the robot magnet are perpendicular, making the fin-like Robots. (b) The actuator magnet and the robot magnet are parallel to each other making the screw-like Robots.

gradient ($\nabla\vec{B}$). Equation (2) represents the torque ($\vec{T}$) experienced by the magnet in a magnetic field ($\vec{B}$). Depending on how the magnet is oriented in the field, either or both of these components could be induced, in this study two different orientations are analyzed.

In the context of the filaments proposed in this work, two configurations called the screw-like and the fin-like robots are proposed and studied. In the case of the fin-like robots, the magnetization vectors of the robot magnet and the actuator magnet are perpendicular to each other (Fig. 3.2.a). In this case, equation (2) where torque is dominant is governing, and the filament experiences torque and rotates to align itself with the actuator magnet. If the filament is placed in a confined environment, for example, a tube, the robot magnet is not able to align itself with the actuator magnet due to the tube-walls reaction moments, the head and tail of the filament hit the barrier and the friction is what drives the filaments forward.

The second case is the case of screw-like robots (Fig. 3.2.b). As discussed in the liter-

ature review, this configuration is quite common in small-scale robotics, where the robot magnet and the actuator magnet have parallel magnetization vector. Similar to the previous case, the robot tries to align itself with the actuator, however, due to the positioning of the vectors, the governing equation is the force equation, rotating the filament synchronously with the actuator. This helical, screw-like motion is the swimming mechanism of this type of filament.

In order to have a working system, the theory must be verified using modeling and experimentation. The choice of optimal dimensions and proof of a functional system can be obtained through simulations as well. As stated in the previous section, the dimensions of the filament are bound by its operation environment, similarly, the dimensions of the robot magnet are bound by the same constraints and also the commercially available micromagnets. In this case, the micromagnets were obtained from Zigmyster (Toronto, CA) with their strongest dimensionally suitable micromagnet being ($0.7\,\text{mm} \times 0.5\,\text{mm}$) made of NdFeB grade 52 with Ni coating (Fig. 3.4.a).

Having these dimensions means that the optimal dimension for the actuator magnet can be computationally obtained. Keeping in mind that the most commercially available and easiest to work with permanent magnets are cylindrical and made of NdFeB grade 52, the robot and actuator magnet can be modeled in COMSOL to obtain the optimal dimensions of the actuator magnet so the maximum amount of force and torque is applied to the robot magnet. For a distance of 50 mm to replicate the distance of the actuator and robot from each other for clinical application, and initializing the height of the magnet to be 25 mm (approximately 1 inch), the radius of the actuator magnet is optimized.

The graph in Fig. 3.3 shows that, to have the optimal amount of force acting on the robot magnet, the actuator needs to be H = 25 mm and R = 40 mm, which is a very big magnet that is realistically and commercially not available to purchase. Experiments showed that a small 1" diameter $\times$ $\frac{1}{2}$" thickness ($25.4\,\text{mm} \times 12.7\,\text{mm}$) magnet made of NdFeB, grade N52, where the surface field of the magnet is 5233 Gauss ($0.5233\,\text{T}$), B_{rmax} is 14,800 Gauss ($1.48\,\text{T}$) and B_{Hmax} is 52 MGOe ($416\,\text{kJ/m}^3$) induces enough forces and torques on the robot magnet to facilitate propulsion, having this small system ensures the safety of usage in clinical settings. Fig 3.5.a shows the robot and actuator magnets which are modelled in MATLAB.

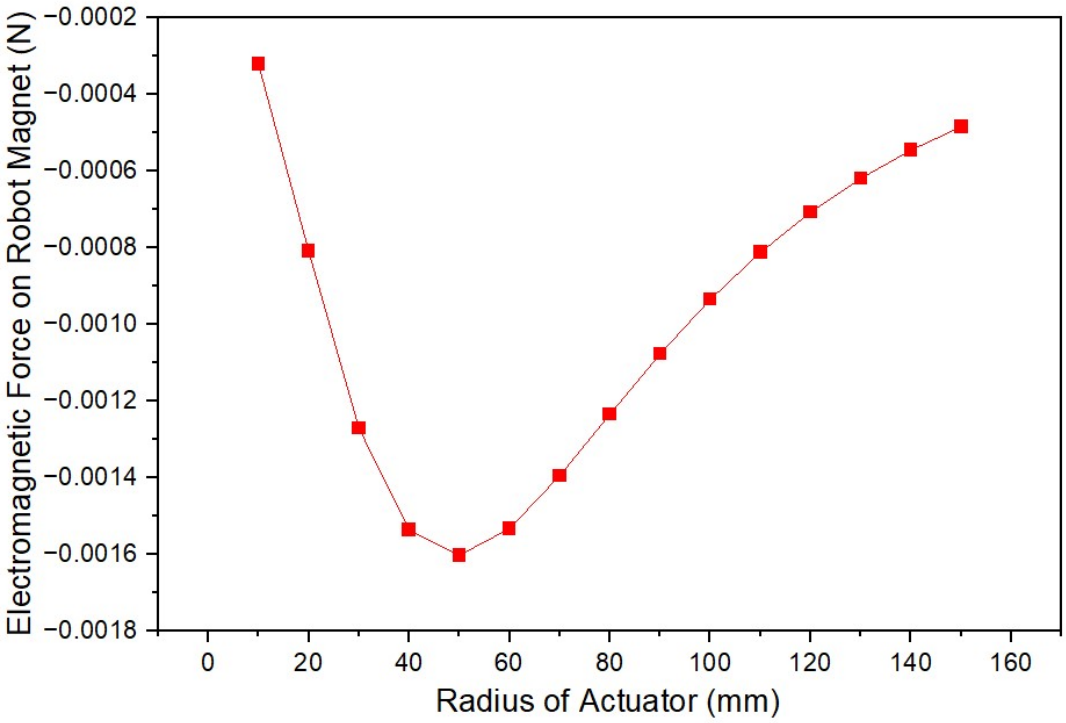


Figure 3.3: Optimization of the Radius of the magnet for a height of 25 mm

If we consider a cylindrical body with the circular radius R and semi-length L (Half the height of the cylinder), centered at the origin of a cylindrical coordinate system (ρ, ϕ, z) with its axis aligned along the z-direction (Fig. 3.4.b). The body's uniform magnetization along an arbitrarily chosen magnetization vector $\mathbf{M}$ can be decomposed into longitudinal and transverse components,

$$\mathbf{M} = M_l \hat{z} + M_t \hat{\rho} \tag{3}$$

In practice, for a magnetizable material, the resultant magnetization generally does not align with the applied field $\mathbf{H}_{\text{ext}}$. Instead, the magnetization vector rotates to minimize its energy based on the material's magnetic susceptibility and the body's demagnetization factors, as detailed by the Stoner-Wohlfarth model [55]. Generally, the magnetization $\mathbf{M}$ relates to the magnetic field $\mathbf{H}$, the magnetic flux density $\mathbf{B}$, and the permeability of

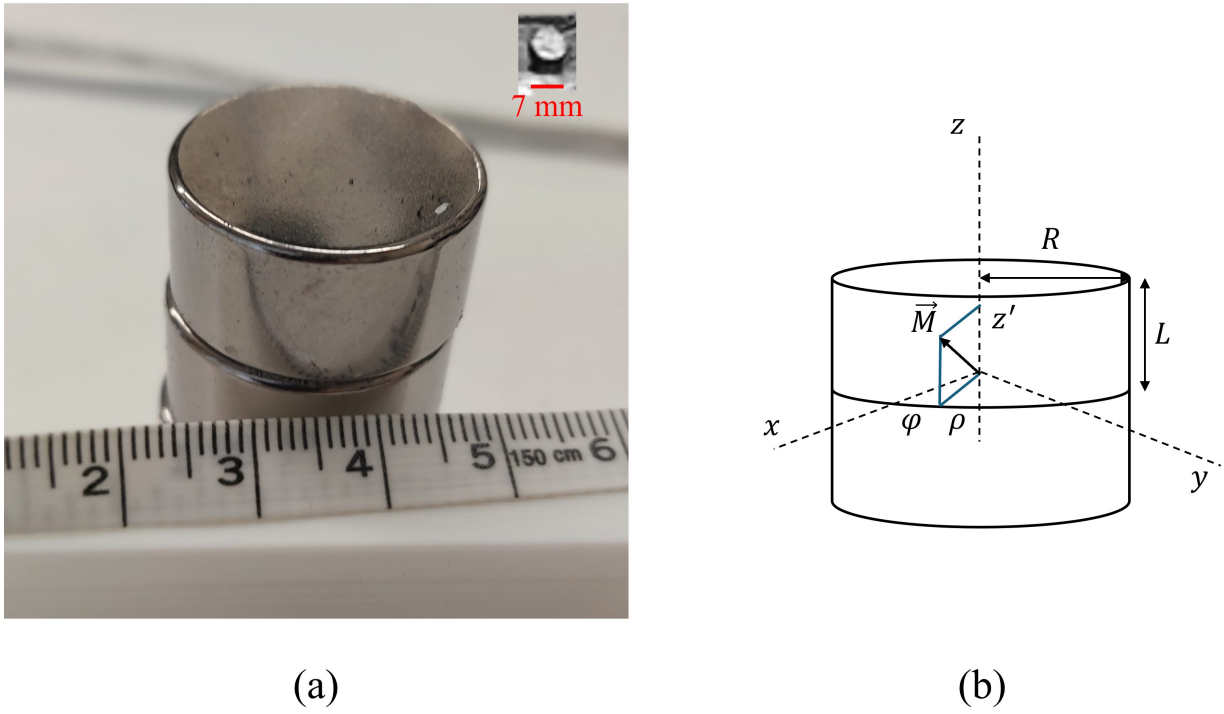


Figure 3.4: (a) Actuator magnet, (b) Robot magnet

vacuum μ_0 through:

$$\mathbf{B} = \mu_0(\mathbf{H} + \mathbf{M}) \tag{4}$$

Callaghan and Maslen initially derived the field equations inside and outside finite cylinder which is magnetized along its longitude, treating it as a series of current loops (i.e. an ideal solenoid) [56]. They considered the total magnetization to be $\mathbf{M} = nI$, with n being the number of turns per unit length and I the current per turn. The magnetic field is calculated directly by elliptic integrals when using the Biot-Savart law. Derby and Olbert revised this derivation, providing a method which is efficient in terms of computation power [57]. They retrieved the field of a current loop in the limit $L \to 0$ and the far-field limit of a point dipole at large distances from the cylinder. Caciagli *et al.* restated these results in

closed form using elliptic integrals presented as

$$B_\rho = \frac{\mu_0 MR}{\pi}\left[a_+ P_1(k_+) - a_- P_1(k_-)\right] \tag{5}$$

$$B_z = \frac{\mu_0 MR}{\pi(\rho+R)}\left[\frac{b_+}{P_2(k_+)} - \frac{b_-}{P_2(k_-)}\right] \tag{6}$$

where B_ρ and B_z are respectively the radial and axial components of the magnetic flux density, $\mathbf{M}$ is the total magnetization, μ_0 is the permeability of free space, R is the radius of the current loop, L is the half-length of the cylinder and ρ is the radial distance from the axis of the cylinder to the point where the magnetic field is being calculated [58]. Two auxiliary functions are defined as:

$$P_1(k) = K - \frac{k^2}{2}(K - E) \tag{7}$$

$$P_2(k) = -\frac{c}{1-c^2}P - K\left(\frac{1}{1-c^2} - \frac{c^2}{1-c^2}P - K\right) \tag{8}$$

The symbols K, E, and P denote the complete elliptic integrals of the first, second, and third kinds, respectively:

$$K = K(\sqrt{1-k^2}) = \int_0^{\pi/2} \frac{d\theta}{\sqrt{1-(1-k^2)\sin^2\theta}} \tag{9}$$

$$E = E(\sqrt{1-k^2}) = \int_0^{\pi/2} \sqrt{1-(1-k^2)\sin^2\theta}\, d\theta \tag{10}$$

$$P = P(1-c^2, \sqrt{1-k^2}) = \int_0^{\pi/2} \frac{d\theta}{(1-(1-c^2)\sin^2\theta)\sqrt{1-(1-k^2)\sin^2\theta}} \tag{11}$$

Also the following shorthand notations are used:

$$n_\pm = z \pm L \tag{12}$$

$$a_\pm = \sqrt{n_\pm^2 + (\rho+R)^2} \tag{13}$$

$$b_\pm = n_\pm a_\pm \tag{14}$$

$$c = \frac{\rho - R}{\rho + R} \tag{15}$$

$$k_{\pm}^2 = \frac{n_{\pm}^2 + (\rho - R)^2}{n_{\pm}^2 + (\rho + R)^2} \tag{16}$$

where n is the number of turns per unit length and z is the axial distance from the center of the current loop. a, b, c and k are merely notations to summarize the final equations and hold no physical meaning.

The magnetic field magnitude B is then computed as the square root of the sum of the squares of B_{qt} and B_{zt}:

$$B = \sqrt{B_{qt}^2 + B_{zt}^2} \tag{17}$$

Since the material and the dimensions of the actuator magnet are chosen, the magnetic field can be modelled in MATLAB.

As shown in Fig. 3.5.a, the magnetic field decreases as the distance from the surface increases. Consequently, the torque induced on the robot magnet follows a similar trend, as it is derived from the magnetic field (Fig. 3.5.c). However, the force reaches its maximum at a certain distance from the magnet (Fig. 3.5.d), behaving in accordance with the field gradient (Fig. 3.5.b). These maximum points are located very close to the magnet, but experiments showed that in further realistic distances, locomotion still occurs.

The component of time could also be added to the MATLAB algorithm to observe the swimming mechanism of the robot. This calculation is only possible in the case of the screw-like robots, where the motion in independent of the boundaries. Also in this case the equations are made for equilibrium and there is no consideration of moment of inertia for simplification purposes.

Magnetic field B is calculated on the body frame of the actuator, therefore, the transformation of the magnetic field to a reference frame is accomplished using rotation matrix $R_y(\theta)$. This matrix defines the rotations and translations along the y-axis, defined as:

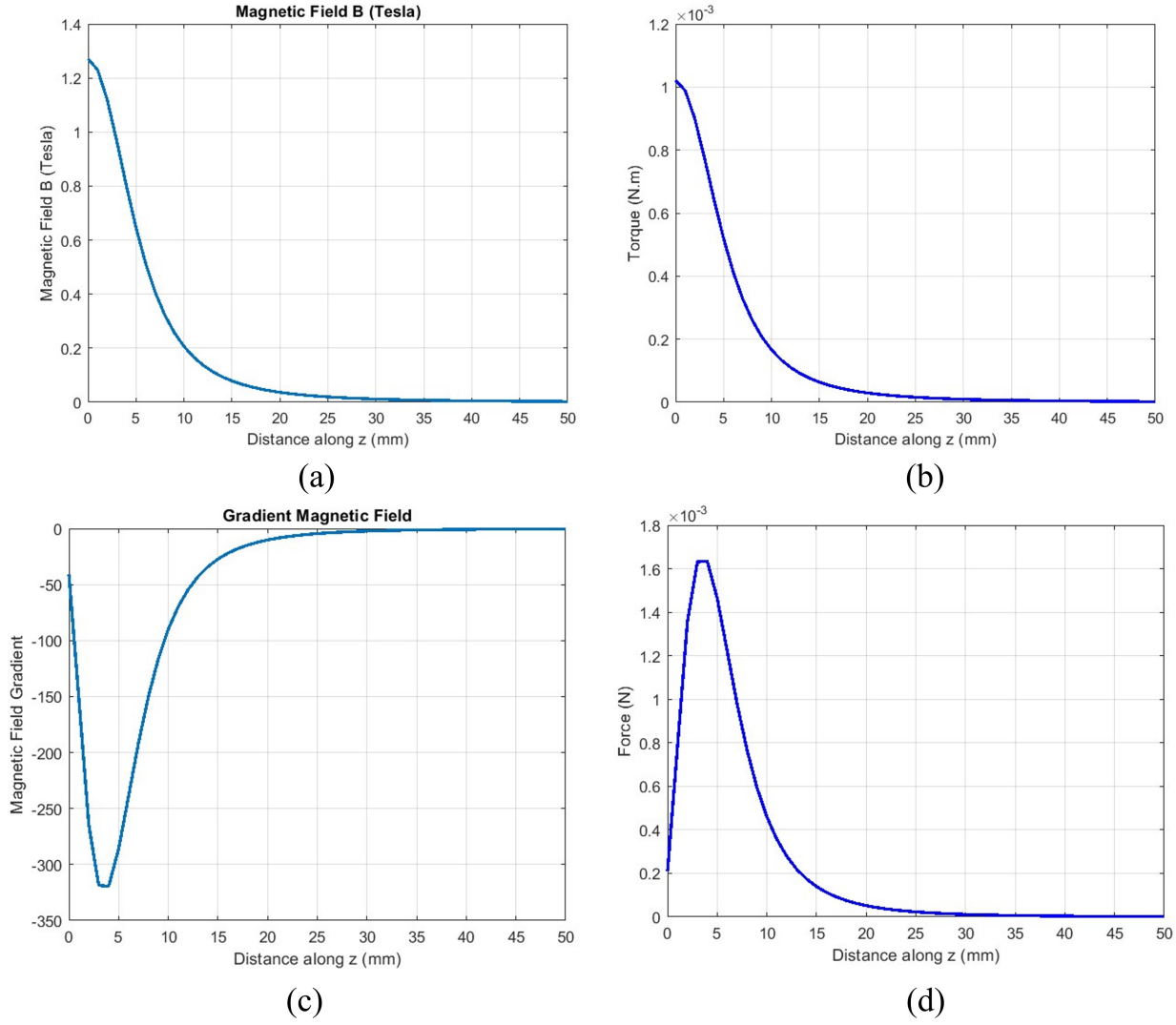


Figure 3.5: (a) The magnetic field produced by the cylindrical actuator magnet along its center-line (b) Torque (Nm) induced on fin-like robot at t=0s over distance from actuator magnet(c) Gradient of the magnetic field produced by the cylindrical actuator magnet along its center-line (d) Force (N) induced on screw-like robot at t=0s over distance from actuator magnet

$$R_y(\theta) = \begin{pmatrix} \cos(\theta) & 0 & \sin(\theta) & 0 \\ 0 & 1 & 0 & 0 \\ -\sin(\theta) & 0 & \cos(\theta) & 0 \\ 0 & 0 & 0 & 1 \end{pmatrix} \tag{18}$$

The gradients of the magnetic field components are calculated using central differences for each time step:

$$\frac{dB}{dx} \approx \frac{B(x + \Delta x) - B(x)}{\Delta x} \tag{19}$$

The properties of the robot magnet are also defined in the robot's body frame, which are then converted to the reference frame using the same transformation matrix, however in this case the rotation degree is unknown. Table 3.1 shows the symbols and variables used in the system of equations and unknowns.

The field and gradient is then used to compute the forces on a robot magnet by solving a system of equations derived from the relationships between the magnetic field gradients and the forces. The equations for the forces F_x, F_y, and F_z are given by:

$$F_x = m_x \frac{\partial B_x}{\partial x} V_m \tag{20}$$

$$F_y = m_y \frac{\partial B_y}{\partial y} V_m \tag{21}$$

$$F_z = m_z \frac{\partial B_z}{\partial z} V_m \tag{22}$$

Then a system of equations and unknowns is used to see if the robot follows the rotation of the actuator at equilibrium.

$$m_x \frac{\partial B_x}{\partial x} V_m - F_x = 0 \tag{23}$$

$$m_y \frac{\partial B_y}{\partial y} V_m - F_y = 0 \tag{24}$$

Symbol	Description	Unknown
ω_2	Rotational velocity of robot (rad/s)	Yes
V_{2x}	Linear velocity of robot in the x direction (m/s)	Yes
V_{2y}	Linear velocity of robot in the y direction (m/s)	Yes
V_{2z}	Linear velocity of robot in the z direction (m/s)	Yes
F_x	Force in the x direction (N)	Yes
F_y	Force in the y direction (N)	Yes
F_z	Force in the z direction (N)	Yes
$B_{x_{\text{act}}}$	Magnetic field component in the x direction (T)	No
$B_{y_{\text{act}}}$	Magnetic field component in the y direction (T)	No
$B_{z_{\text{act}}}$	Magnetic field component in the z direction (T)	No
m	Mass of robot (kg)	No
V_m	Volume of robot (m^3)	No
$m_{x_{\text{rob}}}$	Magnetic moment of robot in the x direction ($\text{A}\cdot\text{m}^2$)	No
$m_{y_{\text{rob}}}$	Magnetic moment of robot in the y direction ($\text{A}\cdot\text{m}^2$)	No
$m_{z_{\text{rob}}}$	Magnetic moment of robot in the z direction ($\text{A}\cdot\text{m}^2$)	No

Table 3.1: Symbolic Variables, Descriptions, and Unknowns

$$m_z \frac{\partial B_z}{\partial z} V_m - F_z = 0 \tag{25}$$

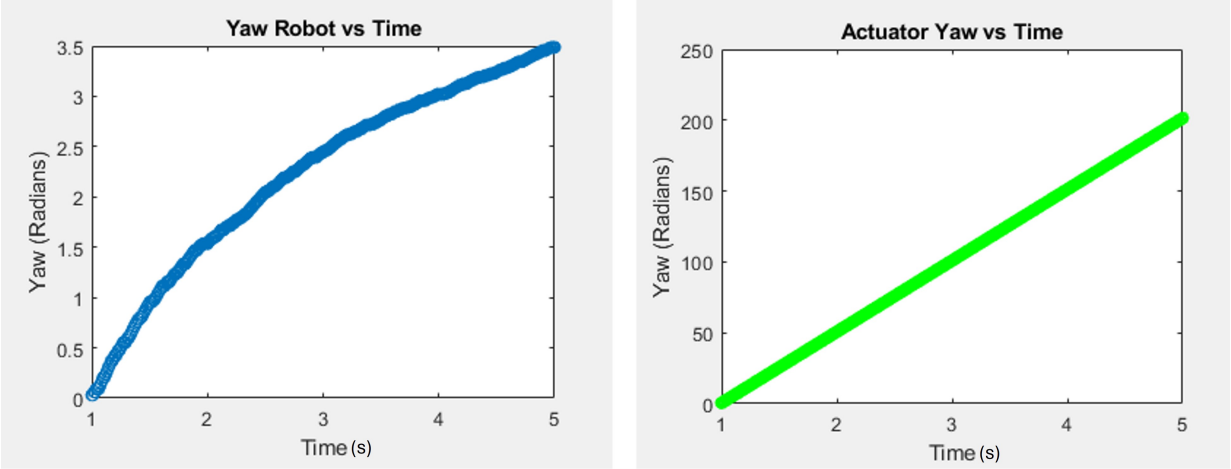
$$m \frac{V_{2x}}{t} - F_x = 0 \tag{26}$$

$$m \frac{V_{2y}}{t} - F_y = 0 \tag{27}$$

$$m \frac{V_{2z}}{t} - F_z = 0 \tag{28}$$

$$m \cdot 10 - F_x = 0 \tag{29}$$

The output of this modelling is demonstrated in Fig. 3.6 showing that as the actuator magnet rotates, the robot magnet rotates with it. The errors between the graphs are due to the simplifying assumptions, i.e. neglecting the effects of moment of inertia and phase delay.



(a) (b)

Figure 3.6: Comparison of the rotation of the robot vs. the rotation of the actuator magnet

3.2 Mechanical Properties and Hydrodynamics

The shape of the filament in this study is a simple cuboid, with the dimensions being chosen based on the application. As the purpose of the robot is drug delivery, the ideal scenario entails the maximum amount of drug which can be loaded, with the targeted organs limiting the dimension. As Fig. 3.1.e shows, the filaments proposed in this case have a square cross-section of 1 mm × 1 mm, this design choice is based on the fact that the smallest confinement in the human urinary tract is in the ureter having the diameter of approximately 4 mm to 5 mm. The length of the robot in this case is 12 mm, which is chosen based on the organ model in test. The dimensions of the filament could be further optimized based on the specific patient in application.

Having this length in mind, the magnet in the screw-like robots and the fish-like robots could be placed anywhere in the length of the filament, which means that the swimming motion of the robot slightly changes in each case. If the filament is assumed to be a flexible beam, and the fluid surrounding the robot to act as simple supports, then Fig. 3.7 approximates the deflection of each type of robot in different magnet placements according

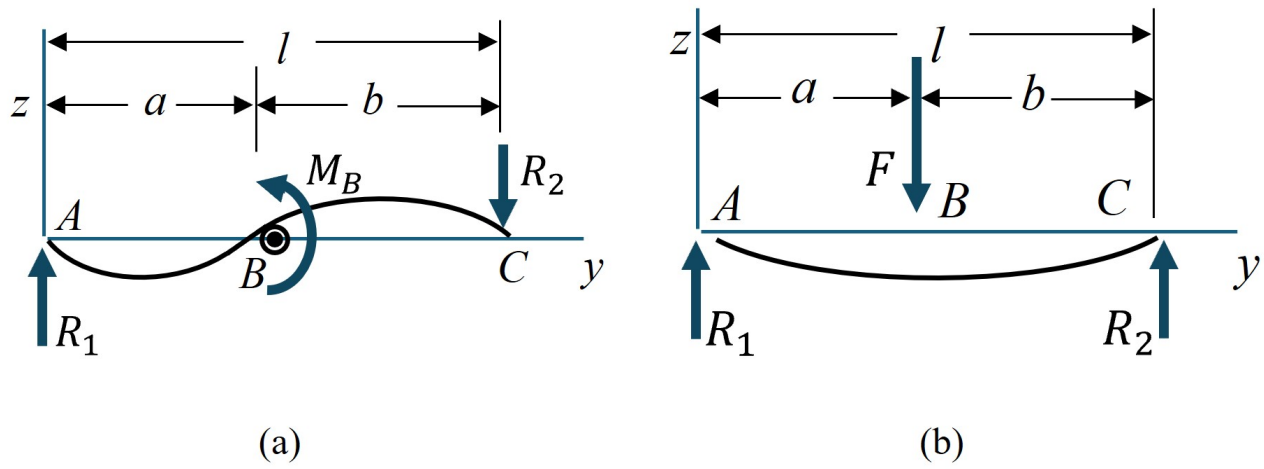


Figure 3.7: (a) Deflection of a beam due to applied moment (b) Deflection of a beam due to applied point force

to the following equations.

The deflection of a simply supported beam when force is applied to it, like the case of the screw-like robots, is calculated as

$$y_{AB} = \frac{F_{bx}}{6EI} \cdot \frac{1}{(x^2 + b^2 - l^2)},$$
$$y_{BC} = \frac{F_a(l-x)}{6EI} \cdot \frac{1}{(x^2 + a^2 - 2lx)} \tag{30}$$

where y_{AB} is the deflection at point B due to the load F_{bx} where x is the horizontal distance from the coordinates' origin, E is the modulus of elasticity of the material and I is the moment of inertia of the beam's cross-section (Fig. 3.7.b).

In the case of the fin-like robots, the deflection can be replicated by a simply supported beam under moment load as per the following

$$y_{AB} = \frac{M_B x}{6EI} \cdot \frac{1}{(x^2 + 3a^2 - 6al + 2l^2)}$$

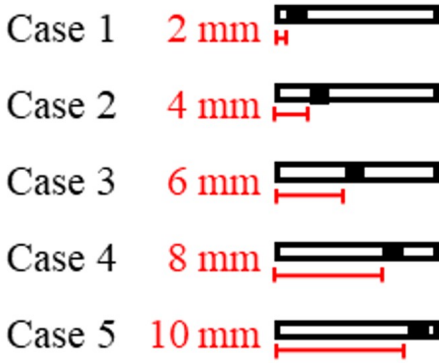
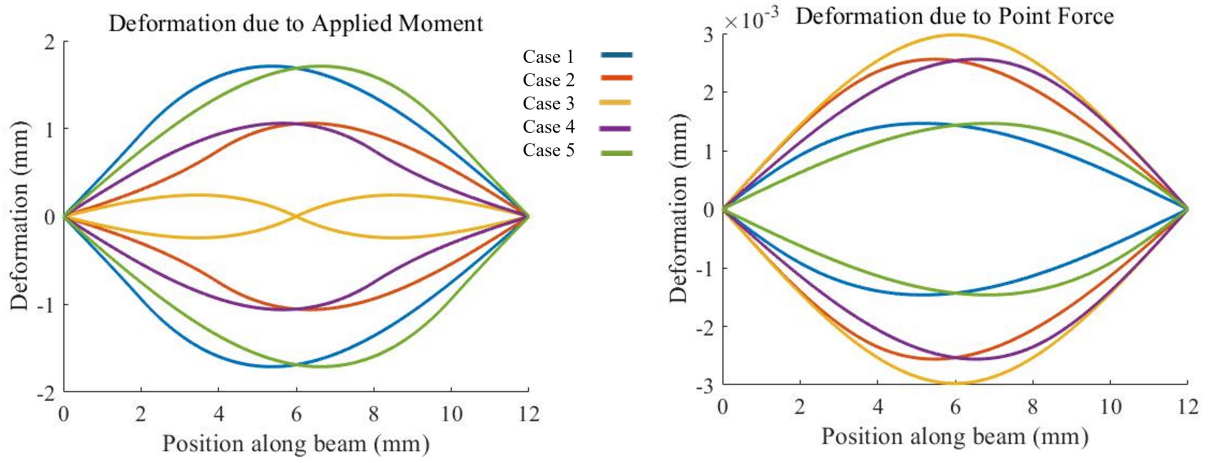


Figure 3.8: The 5 cases of magnet placement studied in this work for frequency response and deformation for both the fin-like and the screw-like robots. The head of the robot is indicated by the black square/ robot magnet in this figure. Assuming the swimming direction for all five cases to be to the left, cases 1 and 2 have the head of the robot moving in that direction and cases 4 and 5 have the tail moving first.

$$y_{BC} = \frac{M_B}{6EI} \cdot \frac{1}{(x^3 - 3lx^2 + x(2l^2 + 3a^2) - 3a^2l)} \tag{31}$$

where y_{AB} is the deflection at point B due to the moment M_B acting at B, x is the horizontal distance from the coordinates' origin, E is the modulus of elasticity of the material and I is the moment of inertia of the beam's cross-section (Fig. 3.7.a).

It should be mentioned here that in addition to analyzing the effect of the orientation of the robot magnet on the filaments' response, the effect of the placement of the magnet along the filament was also studied for 5 different magnet positions (Fig. 3.8), the difference between case 1 and 5, and 2 and 4 being the what part of the robot is closer to the actuator magnet. In cases 1 and 2 the filament is assumed to swim head-first, with the robot magnet being closer to the actuator magnet, and in cases 4 and 5 the robot swims tail-first where the tail with no magnet is closer to the actuator magnet. Using the forces and torques calculated in the previous section, the deformation of the screw-like and fin-like robots under static force and moment load is analyzed per equation 3 and 4, considering the modulus of elasticity to be 120 kPa as stated in [59]. Fig. 3.9 shows the results of this modeling, showing that we expect 1000 times more deformation in the fin-like robots.



(a) (b)

Figure 3.9: (a) Deformation of fin-like robots due to applied moment (b) Deformation of screw-like robots due to applied point force

The previously stated equations show the structural behavior of the filaments, however, it should be kept in mind that the working environment of this system is the urinary tract and the filaments are working in fluid. In order to have a general idea of how the filaments work in fluids, Reynolds number can be used. Reynolds number is a dimensionless parameter that characterizes the relation of inertial forces (ρu^2)to viscous forces ($\mu u/L$) within a fluid flow. When it exceeds 1, inertial forces dominate, leading to a transition from laminar to turbulent flow behavior. This transition signifies a shift from a predictable, well-ordered flow to a more chaotic and unpredictable regime with increased mixing and energy dissipation. In such scenarios, the dominance of inertial forces leads to complex flow patterns and necessitates sophisticated computational approaches for accurate prediction. On the other hand, when Reynolds falls below 1, viscous forces become paramount, dictating the fluid behavior. This regime is mostly present in micro-scale applications, where micro-robots mimic the locomotion of microorganisms. The dominance of viscous forces allows for precise control and manipulation of the fluid environment, and simpler calculations modelling the flow. Reynolds number is calculated as

$$\mathrm{Re} = \frac{\rho \cdot V \cdot L}{\mu} \tag{32}$$

where Re is the Reynolds number, ρ is the density of the fluid, V is the velocity of the fluid, L is the characteristic length, and μ is the dynamic viscosity of the fluid. In this research, Reynolds number can be calculated by assuming that the surrounding fluid is water with the density (ρ) of $1000\,\mathrm{kg/m^3}$, average velocity of the filament (v) is experimentally evaluated to be approximately $5\,\mathrm{mm/s}$, length of filament (L) is 12mm and viscosity of water (μ) is $0.001\,\mathrm{Pa.s}$, giving Reynolds number to be 60.

The calculated Reynolds number in this case shows a non-laminar flow regime. This implies that inertial forces govern the fluid motion surrounding the filament, facilitating its locomotion. A detailed analysis of the full hydrodynamic interactions experienced by the filament would necessitate solving the Navier-Stokes equations, which is beyond the scope of this present research. However, acknowledging this non-laminar behavior is crucial for understanding the dominant forces influencing the filament's movement and for potentially exploring simplified models that capture the essential dynamics.

3.3 Actuation System Overview

3.3.1 Robotic Arm

The experimental setup for the magnetic actuation of flexible filaments includes a robotic arm, demonstrated in Fig. 3.10 which is a 4 DoF system from igus. The controller of the robot houses the computer and motor driver modules that control its movements, which is then linked to iRC, the software provided by the company, allowing the robot to be moved through jogging controls and programs for automated tasks.

Each axis of the robot is moved by a bipolar stepper motor with an encoder, and the feedback of the encoders is also provided to the controller for proper positioning. Furthermore, the robot can be referenced after cold starts so the actual position of the robot and the software act in sync. All the control modules for the robot are in a PLC

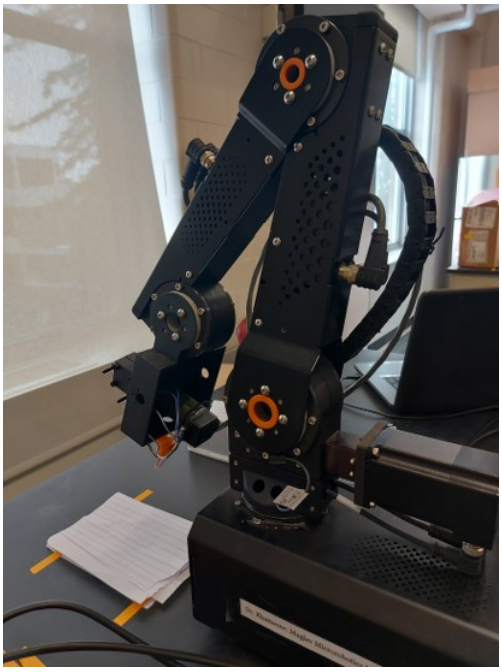


Figure 3.10: Robot Arm from igus

cabinet for ease of use and the entire system is paired with the computer using a CAN to USB adaptor which facilitates communication between the controller area network (CAN) and a computer. As previously stated, this robot arm only provides 4 degrees of freedom, meaning it has 4 stepper motors, controlling the position on the end-effector in the x, y and z direction. However, it can only rotate the end effector around the x-axis, and rotation around y and z is not possible.

In clinical settings, any robotic arm with proper precision and control could be implemented to provide the external actuation required for the soft robots. In this case, using the GUI of iRC helped program the robot arm to move in a repeatable fashion, producing repeatable results for the movement of the filaments. The iRC environment allows the user to define linear paths for the robot arm to follow. Furthermore, the organ models are positioned accordingly for the robot arm to maneuver the rotating magnetic field. In addition, a mounting bracket was designed and 3D printed from resin for the robot arm which provides space for a DC motor and a magnet, components necessary to provide a

rotating magnetic field.

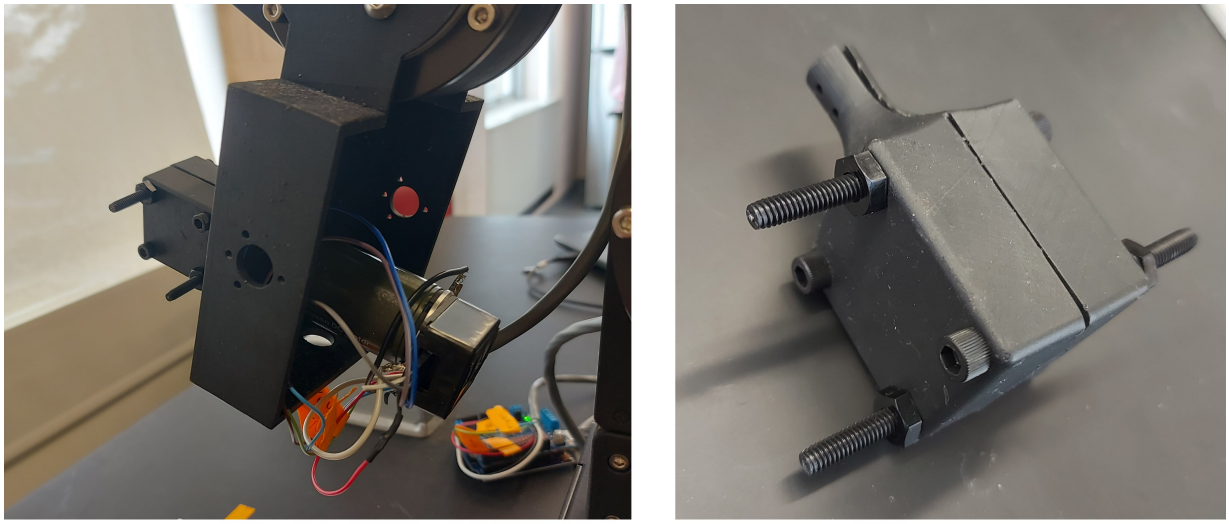
3.3.2 Rotating Magnetic Field

This arm is equipped with a brushed 24-volt DC motor from Maxon Motor, attached to its end-effector i.e. the 3D printed bracket (Fig. 3.11). The motor enables rotational movement of a permanent with dimensions of 1" diameter $\times \frac{1}{2}$" thickness (25.4 mm $\times$ 12.7 mm) made of NdFeB, grade N52, where the surface field of the magnet is 5233 Gauss (0.5233 T), B_{rmax} is 14,800 Gauss (1.48 T) and B_{Hmax} is 52 MGOe (416 kJ/m^3). A specially designed holder, which is also designed and 3D printed, secures the magnet to the DC motor shaft, creating a rotating magnetic field and gradient. The choice of the magnet for this case is through commercial availability, however, in further developed settings, the actuator magnet can be further optimized to have the most fitting magnetic field for each setting.

The important variable, in the case of external actuation, is controlling the speed with which the actuator magnet rotates. For that reason, the DC motor also has an optical encoder (Avago Technologies) which facilitates monitoring the rpm of the system. The rotation set up uses Arduino as a controller, where the input voltage of the motor is entered into the system and the resulting rpm is observed. Using this method, the rotation frequency of the magnetic field can be adjusted by the operator.

3.3.3 Camera and Tracking

In order to collect statistical data from the experiments, the result of the movements from the filaments in each organ model at different frequencies has to be recorded and analyzed. Therefore, a camera from Basler along with a magnifying lens was used to record videos of the soft robots' response to the actuation at each frequency in different organ models. The Basler camera is paired with Pylon Viewer as a software which helps the operator set the exposure time and frame rate of the recordings for precise analysis of the movements. If the frequency of actuation is higher than 16 the frame rate was set to 75 and for lower frequencies, a frame rate of 50 was chosen to ensure high-quality video capture for processing.



(a) (b)

Figure 3.11: (a) DC motor attached to the robot arm's end-effector, (b) Actuator magnet inside a holder

The next step in the experiments is analyzing the locomotion of the filaments in different environments at different frequencies through image analysis. The tool used in this case is called NMTT which is a python script available online (https://github.com/rafamestre/NMTT-nanomicromotor-tracking-tool). This program is an image processing and tracking algorithm which can follow different particles based on the contrast between the pixels of the particle and the background. At each frame of the recording, the relative pixel coordinates from the starting point is saved and transformed to distance by a scale provided by the operator. The analysis was done for each recording, giving a pool of data which was then statistically analyzed.

3.3.4 Anatomical Model of the Urinary Tract

The testing of the filaments was carried out in a 3D-printed urinary tract organ model which was designed based on anatomical features obtained from medical images and re-

constructed into a three-dimensional design printed from flexible, transparent resin. The urinary tract model was segmented from real patient CT scans sourced from The Cancer Imaging Archive. The renal pelvis, ureter, and bladder were segmented individually in Slicer 3D, refined, and hollowed for printing in Meshmixer. The model represents an adult's urinary tract. The resulting renal pelvis model has a height of 41.30 mm and a maximum major calyces length of 43.28 mm. The segmented ureter's diameter ranges from 2.54 to 5.14 mm with a length of 123.3 mm, while the average adult ureter length is 280-320 mm. This was adjusted by adding a silicon tube with an inner diameter of 5/16" to achieve a total length of 280 mm. The bladder measures 75.05 mm in length, 59.10 mm in height, and 65.73 mm in diameter, with wall thicknesses of 0.40 mm for the renal pelvis and ureter, and 3 mm for the bladder. Printing was done using Elastic 50A resin on the Form3 printer. Post-processing involved specific steps, the renal pelvis and ureter underwent a 20-minute ethanol wash, ethanol flushing to remove residual resin, a 30-minute air dry, a 33-minute UV cure, and support removal. The bladder was split in half before printing to prevent cupping, then required a 10-minute ethanol wash, a 20-minute UV cure, and support removal. For the frequency response tests, only half the printed bladder was used to ensure quality of recording the results.

3.4 Robot Fabrication and Kidney Stone Dissolution

The fabrication of these miniature magnetic robots utilizes a biocompatible hydrogel known as Gelatin-methacrylate (GelMA). GelMA, specifically Cellink 5208, is chosen for its favorable properties, including printability, flexibility, cytocompatibility, and the ability to undergo light-induced cross-linking, making it easy to fabricate. A 10% w/v hydrogel solution is prepared by mixing GelMA powder with a sterile solvent, such as deionized water or phosphate-buffered saline (PBS). Then a biocompatible photoinitiator is added to the solution, namely, Lithium phenyl(2,4,6-trimethylbenzoyl)phosphinate (LAP), at a concentration of 4 mg/mL. LAP plays a crucial role in the photocrosslinking process, as it absorbs light at a specific wavelength and initiates the formation of covalent bonds between GelMA macromolecules, ultimately leading to cross-linked hydrogel formation. This is the base solution for the filaments, and for applications in the urinary tract and kidney stone

dissolution, at this step the desired drug would be mixed with the hydrogel solution.

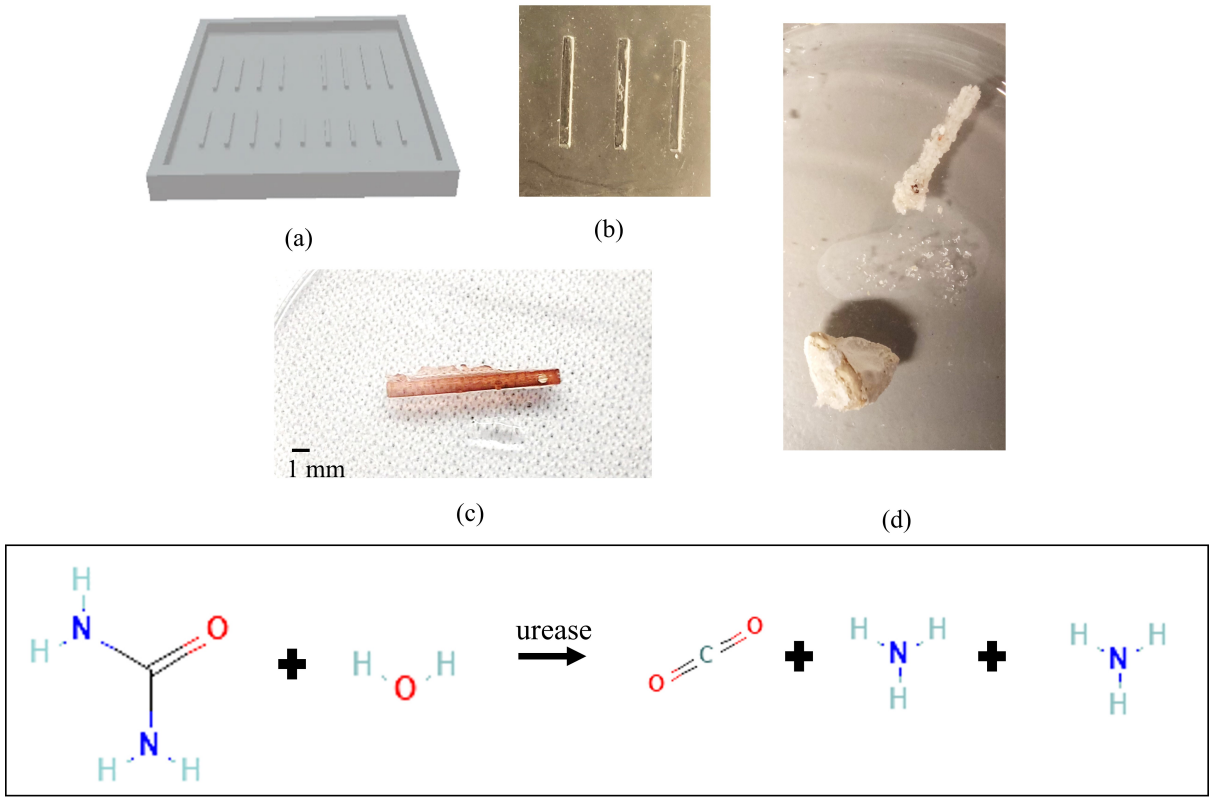


Figure 3.12: (a) Design of the PDMS mold (b) PDMS mold (c) Dyed GelMa filament with embedded permanent magnet (d) GelMa filament and Kidney stone after dissolution experiment (e) The chemical reaction catalysed by urease through which urine becomes basic, in this reaction urea reacts with water resulting in carbon dioxide and ammonia (Molecule figures taken from PubChem)

To have the robots be magnetic and capable of responding to an external magnetic field actuation, permanent micromagnets ($0.7\,\text{mm} \times 0.5\,\text{mm}$) (Zigmyster Magnets, Toronto) are embedded within each filament. These magnets are made of Neodymium Iron Boron (NdFeB), a specific grade (52) is often preferred due to its optimal balance of magnetic

properties. Furthermore, a nickel coating is applied to the NdFeB magnets to enhance their biocompatibility and resistance to corrosion in physiological environments.

The next step involves molding the GelMA solution into the desired filamentous architecture. This is achieved by pipetting the solution into prefabricated PDMS molds and placing the magnet in the desired location and orientation. The filled molds are then exposed to ultraviolet (UV) light with a wavelength of 405 nm for 3 minutes. This exposure triggers the aforementioned photocross-linking process, covalently linking the GelMA chains and transforming the liquid solution into the final, soft filamentous robot structure as depicted in Fig. 3.12.

Finally, to enhance the visibility of the microrobots during manipulation and experimentation, a staining process is employed. One option is to submerge the filaments in a propidium iodide (PI) solution. PI is a fluorescent nucleic acid stain that binds to the cell membrane, rendering the robots a distinct red color. Alternatively, Rhodamine B can be used which makes the filaments pink. This staining facilitates sufficient contrast for the tracking software to be able to process the videos. The final dimensions of the fabricated microrobots are approximately 12 mm in length, with a square cross-section of 1 mm by 1 mm.

The next big step is the choice of the drug for the dissolution of uric acid kidney stones. Urinary pH is a critical factor influencing uric acid stone formation. Lower pH environments cause stone formation, whereas a pH above 7, ideally between 7.0 and 7.2, significantly enhances uric acid dissolution [1]. Traditional basifying agents, such as citrate or bicarbonate, when orally digested are a good solution, but they require continuous administration for extended periods. The approach suggested in this work utilizes urease-loaded filaments to achieve sustained, local enzymatic action within the urinary tract. Urease catalyzes the hydrolysis of urea, a naturally occurring component of urine, into ammonia and carbon dioxide, elevating urinary pH towards a more basic state. This enzymatic approach offers the advantage of a continuous and localized effect, potentially reducing the dosage and side effects associated with conventional basifying agents.

Choosing urease as the desired drug adds the preparation of the enzyme to the process of fabrication. Urease from Canavalia ensiformis (Jack bean) Type III, in powder form

(15,000-50,000 units/g) comes in pulverized form which needs to be mixed with PBS solution for activation. Different concentrations of urease in PBS were tested to achieve the optimal maximum amount which is possible in fabrication, then the solution of urease is added to the GelMa and cured under UV light. The urease concentrations in the GelMa solution were set at 3 mg/mL (n=23), 5 mg/mL (n=45), and 8 mg/mL (n=14), with controls consisting of GelMa filaments without urease (n=14).

In order to have a comprehensive study, bicarbonate and citrate were also tested to gauge whether they would be viable options in the dissolution. However, in the case of bicarbonate, very big amounts of the powder needed to be mixed in with GelMa which made the filaments lose their structural integrity. Furthermore, when the filament is released into the urinary tract, the bicarbonate was released quickly and washed out through the flow. The same principle applies to citrate, although citrate does not mix with GelMa at all, making it impossible to run experiments. This proves urease to be the most efficient option as it is not a drug which gets released, but is an enzyme which is embedded in the filament and stays active, catalyzing the acidifying reaction over long periods of time.

To replicate the physiological conditions of kidney stones in human body, a recipe was followed to make synthetic human urine [60]. This recipe includes mixing all the components commonly found in human urine, namely calcium chloride, magnesium chloride, sodium chloride, sodium sulfate, tridosium citrate dihydrate, monopotassiumphosphate, potassium chloride, ammonium chloride, urea and creatine were added to milliQ water and mixed. Afterward, the synthetic urine needed to be filtered or sterilized to be used in uric acid stone dissolution experiments.

3.5 Experimental Design

3.5.1 Locomotion and frequency response

In this study, we aimed to investigate the optimal location and orientation of the magnet in a basic filament soft robot to enhance its locomotion capabilities within the urinary tract environment.

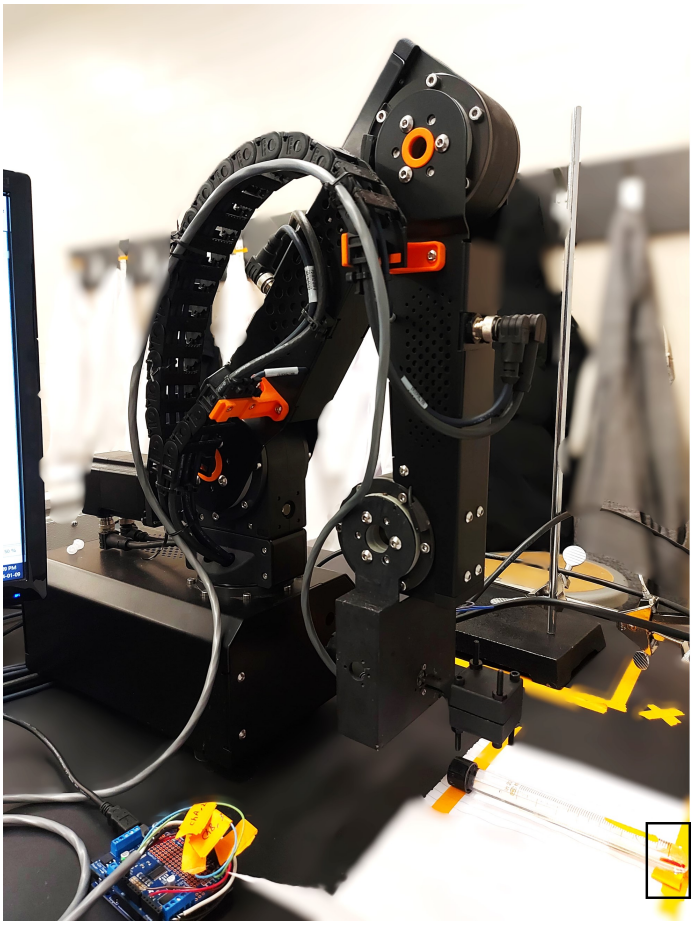


Figure 3.13: Filaments in the ureter model in a tube of water indicated in , placed under actuation and being recorded

The experimental setup was designed to test ten different configurations of soft robotic filaments under controlled conditions. These 10 configurations include the 5 different magnet placements from head to tail in the screw-like and the fin-like robots. Each configuration varied the position and orientation of the magnet embedded within the filament structure to explore how these factors influence the robot's locomotion when subjected to a magnetic field in the ureter, bladder and renal pelvis.

The filaments were tested in a tubular environment designed to simulate the conditions

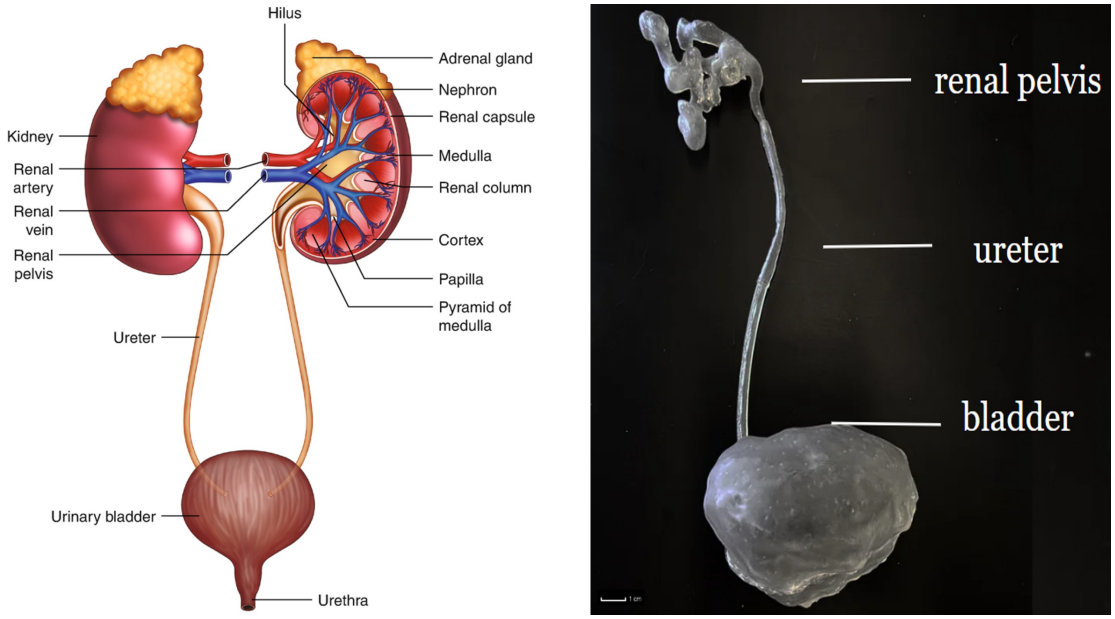


Figure 3.14: Schematic of the human urinary tract [61] (© 2023 Springer Nature) (a) and the full 3D-printed organ model used for actuation experiments in this research (b)

these robots encounter in the ureter. Each configuration was tested using a detailed procedure with the tube's diameter being similar to the ureter (4 mm). For each configuration, ten robots were tested. Each robot was subjected to magnetic actuation at varying frequencies, and a high-speed camera from Basler was used to record the robots' motion. For each frequency, five video recordings were made for each robot, ensuring a comprehensive dataset. The recordings were then analyzed using the Nano-micromotor Tracking Tool (NMTT) which enables the frequency analysis of the filaments.

Following the collection of data for the simple environment of ureter, the filaments were also tested in the bladder where there is no confinement for the robots, which is important in the case of fin-like robots. According to the results of the tests from the ureter, the most optimum location and orientation of the magnet was chosen, and that specific placement for both configurations was tested in the bladder and videos were recorded to be analyzed for acquirement of quantitative data for the frequency response of the robots.

In the last step, both configurations were also tested in the renal pelvis, which includes

spaces with and without confinement to observe how the robots behave in mixed environments. However, in this case the results were only analyzed qualitatively as the complexity of the environment does not allow for proper tracking and actuation of the filaments with the current set up, with the main obstacle being the fact that the robot arm has only 4 degrees of freedom.

3.5.2 Dissolution of Kidney Stones

The materials for the experiments include the filtered and sterile synthetic urine, the GelMa solution mixed with the medicine, the medicine in question, uric acid kidney stones from a laboratory, test vials, and an agitation device which replicates the fluid flow and the movement of a person.

The aim of these experiments was to observe the efficacy of localized chemolysis of uric acid kidney stones. In medicine, this process is achieved through the administration of bicarbonate or citrate. Initially, the attempt was to mix in a soluble amount of these drugs in powder form into the GelMa, however, citrate proved to be completely incompatible with GelMa, and when mixed, the solution fully separated. After this case, bicarbonate was tested in different concentrations which also did not mix very well, needed to be mixed in big amounts to be effective which made the filament brittle and finally, they were not active in the hydrogel, and they were released from the GelMa in a few hours.

Each filament, fabricated as described in section 3.3, was placed into a test tube containing a single uric acid kidney stone which was acquired from a laboratory (initial masses ranging from 0.0019 to 0.0541 g) and 10 mL of sterile-filtered synthetic human urine. The pH of the synthetic urine was adjusted to pH 5.5 with 0.1 M HCl solution to replicate the optimum conditions of uric acid stone formation. Initial pH measurements were recorded using a pH meter, and the samples were then placed on a rotating mixer to agitate continuously for 24 hours. Following this period, the pH of each test tube was recorded again using the pH meter and the synthetic urine medium was replaced. This process was repeated for four consecutive days. At the end of the experimental duration, the kidney stones were removed from the test tubes, dried in an incubator at 60°C for 72 hours, and

their final masses were recorded. The percent change in kidney stone mass was calculated by comparing the initial and final masses.

Chapter 4

Experimental Results and Discussion

4.1 Locomotion and Frequency Response

The results of the experiments provided valuable insights into the locomotion capabilities of the different robot filament configurations within different settings. The following image series (Fig. 4.1 – 4.5) show the movement of the filaments for 8 Hz actuation frequency in a tubular environment, which helps visualize the motion of the different robots and in different cases of magnet placement.

As the image series show, the screw-like robots move in perfect helical rotary motion and are very stable, following the actuator magnet at all times. On the other hand, the fin-like robots do not behave as predictably. As the magnet embedded in them turns to align with the actuator magnet, the filament's body hits the tube's wall and restricts its motion. This wall interaction seems to be what drives the fin-like robots forward. However, at 8 Hz, there is no significant difference between the movements in most cases. The discrepancy happens in Fig. 4.5 where the fin-like robot has almost no movement, this is because experiments showed that the fin-like robots are more self-propelled, choosing their swimming direction and only moving in that direction, which in most cases is head first. However, in higher frequencies, there is a great difference between the different robots' responses. (Fig. 4.6, figures obtained from the tracking algorithm)

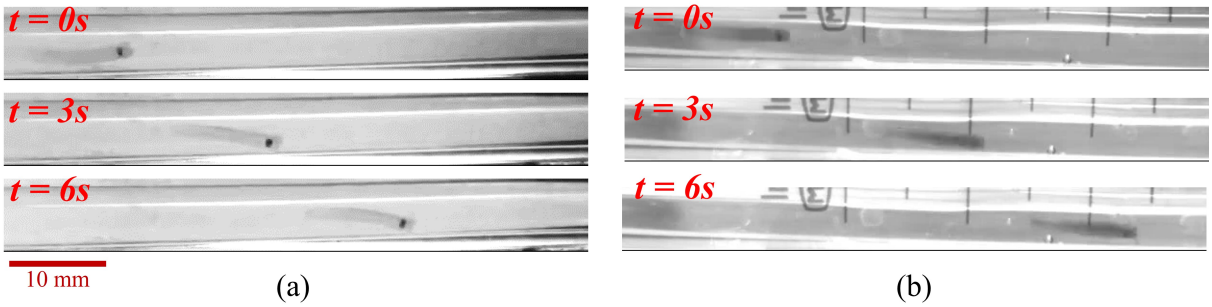


Figure 4.1: (a) Movement of the fin-like robots inside a tubular environment with the magnet embedded as case 1 over a time period of 6 (s), (b) Movement of the screw-like robots inside a tubular environment with the magnet embedded as case 1 over a time period of 6 (s)

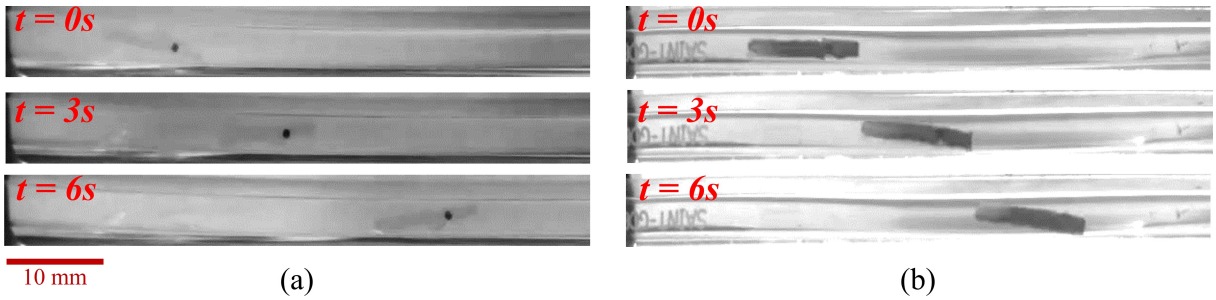


Figure 4.2: (a) Movement of the fin-like robots inside a tubular environment with the magnet embedded as case 2 over a time period of 6 (s), (b) Movement of the screw-like robots inside a tubular environment with the magnet embedded as case 2 over a time period of 6 (s)

The linear velocities of the robot filaments as a function of the actuation frequency are illustrated in Fig. 4.7, revealing distinct behaviors between the screw-like and fin-like robots.

The screw-like robots exhibited a consistent and controlled motion, easily maneuvering forward and backward within the tube at all frequencies. The linear velocity of these

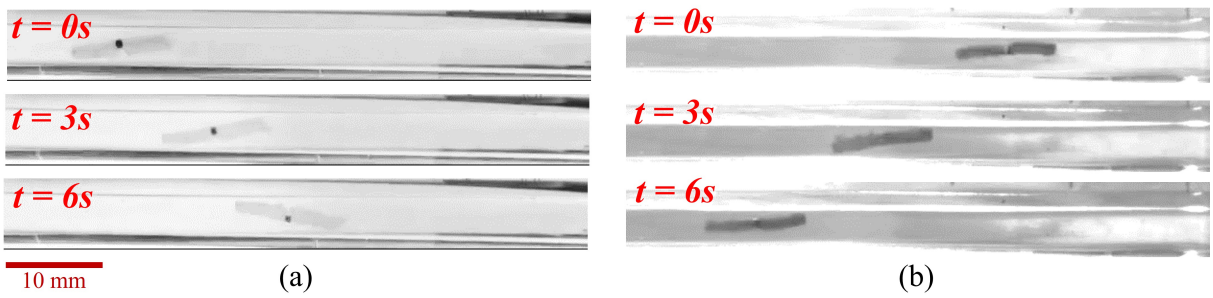


Figure 4.3: (a) Movement of the fin-like robots inside a tubular environment with the magnet embedded as case 3 over a time period of 6 (s), (b) Movement of the screw-like robots inside a tubular environment with the magnet embedded as case 3 over a time period of 6 (s)

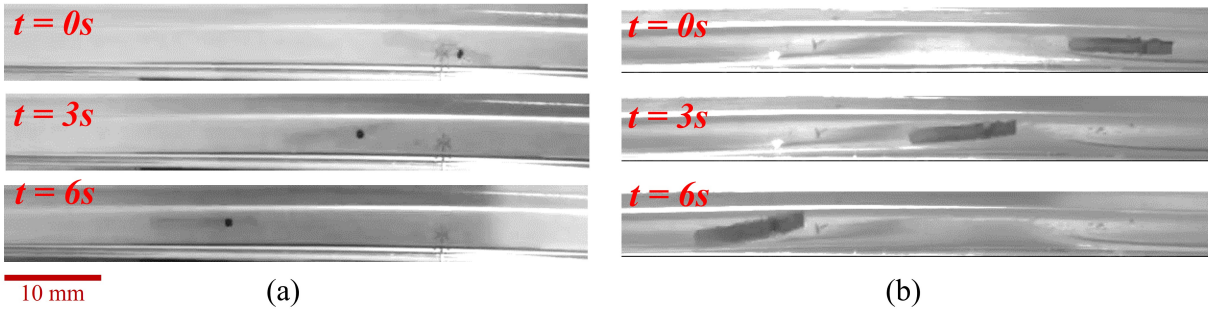


Figure 4.4: (a) Movement of the fin-like robots inside a tubular environment with the magnet embedded as case 4 over a time period of 6 (s), (b) Movement of the screw-like robots inside a tubular environment with the magnet embedded as case 4 over a time period of 6 (s)

robots leveled off above 10 Hz, indicating a saturation point, beyond which increasing the frequency did not significantly enhance the speed. This behavior suggests that screw-like robots are well-suited for applications requiring precise control and bi-directional movement.

The fin-like robots demonstrated significantly higher average velocities, reaching up

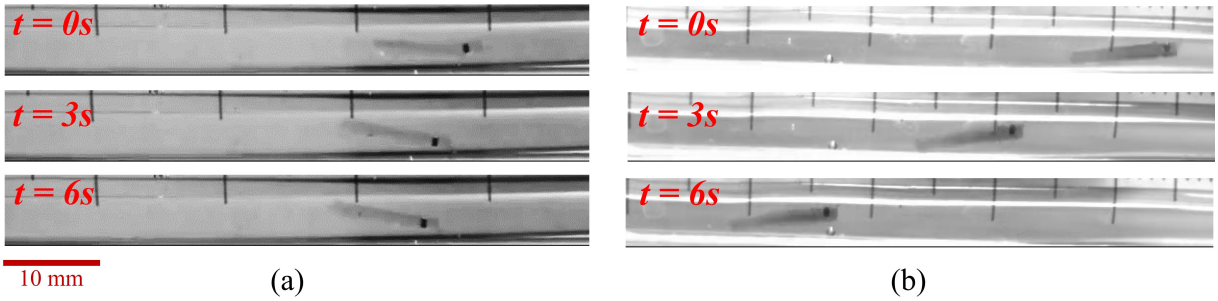


Figure 4.5: (a) Movement of the fin-like robots inside a tubular environment with the magnet embedded as case 5 over a time period of 6 (s), (b) Movement of the screw-like robots inside a tubular environment with the magnet embedded as case 5 over a time period of 6 (s)

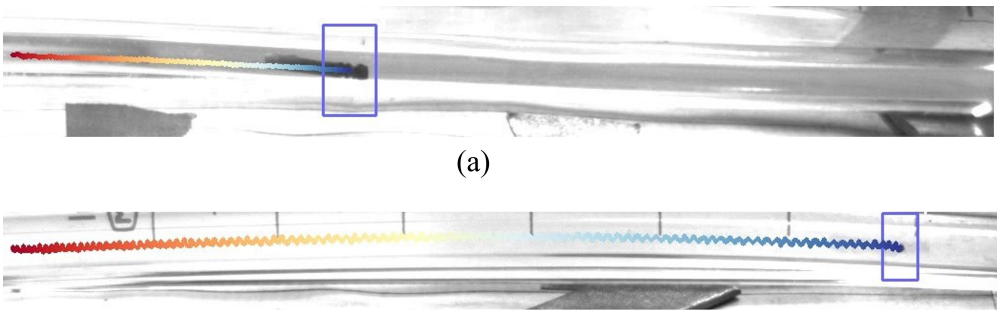
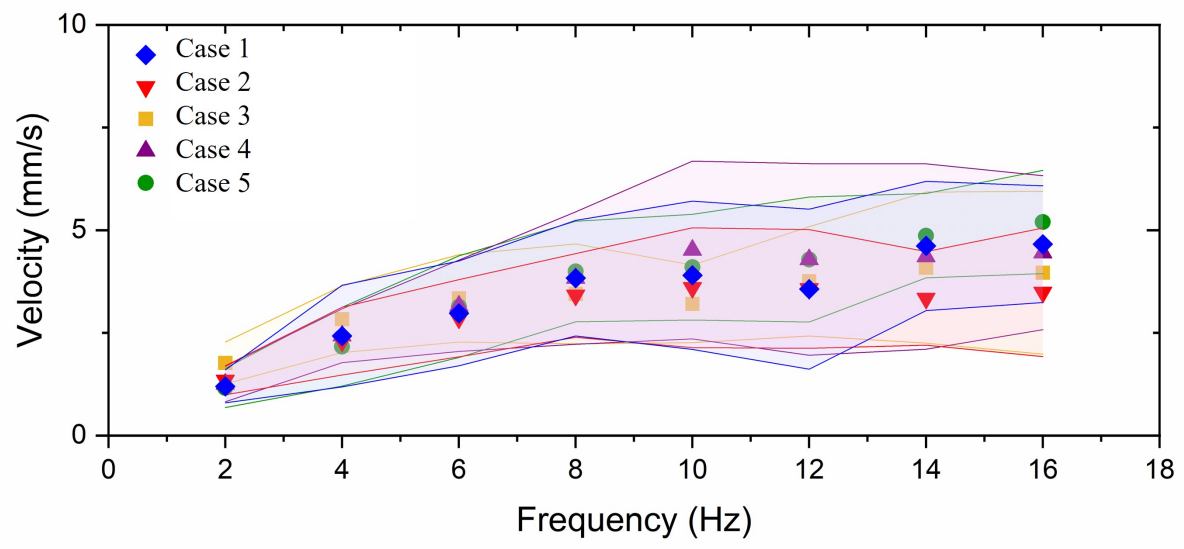
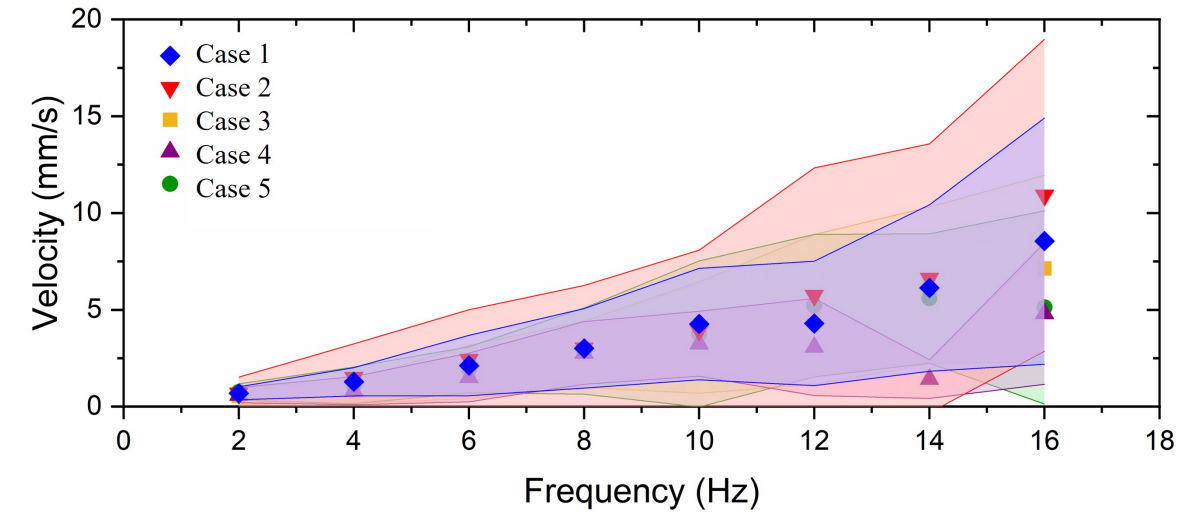


Figure 4.6: (a) Movement of the screw-like robots inside a tubular environment at 16 Hz , (b) Movement of the fin-like robots inside a tubular environment at 16 Hz

to 18 mm/s, compared to the 6 mm/s observed for the screw-like robots. This suggests that the fin-like configuration is highly effective in generating rapid locomotion within confined spaces. However, these robots were more challenging to control, particularly at higher frequencies where the interaction with the tube walls became more pronounced. This increased difficulty in control was attributed to the inherent crawling motion mechanism, which relies on friction with the surrounding surface. Interestingly, the variations in



(a)



(b)

Figure 4.7: Frequency response of the screw-like (a) and the fin-like (b) robots

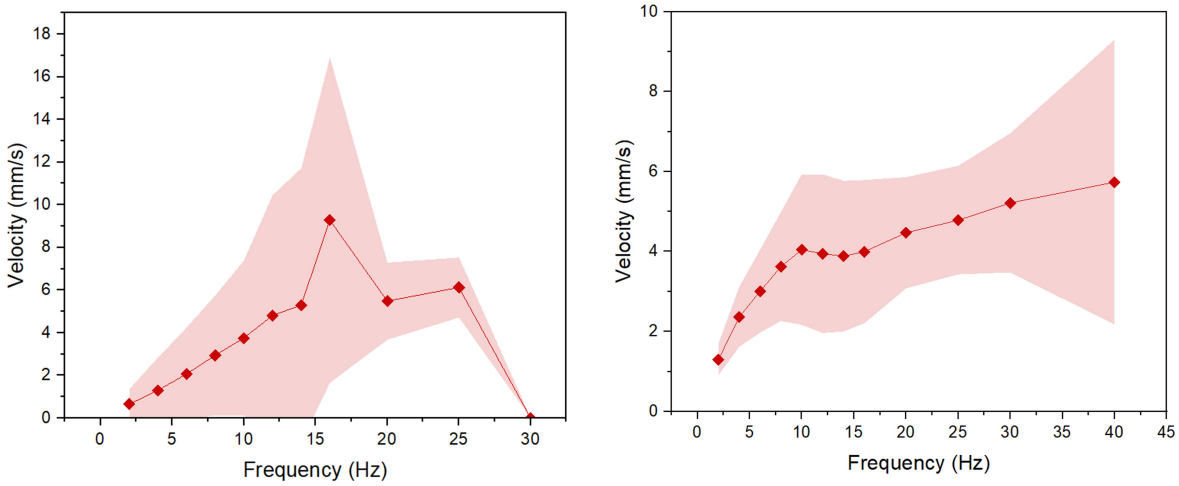


Figure 4.8: Frequency response of the fin-like (a) and the screw-like (b) robots for case 2 and 4 combined at higher frequencies

magnet location and orientation had no significant impact on the frequency response and resulting average velocity of the robot filaments. This observation implies that the primary factors influencing locomotion efficiency are the robot's structural design and the actuation frequency rather than the precise positioning of the magnet. Therefore, in further testings regarding the location of magnet, only case 2 and 4 were considered with no difference in regard to backward and forward motion. Later, higher frequencies were also tested in the ureter to observe a possible plateau or step out in the response of the filaments. Fig. 4.8 shows the results which indicate that the screw-like robots continue to respond well in higher frequencies, however the difference in their response to frequencies higher than 10 Hz is insignificant. However, the fin-like robots stop responding to the actuation at frequencies between 20 and 30 Hz with each one stepping out of the locomotion at a slightly different frequency.

The findings of the tests in the tubular environment were verified when the filaments were tested in the bladder, which does not provide any confinement. As the previous tests showed that the placement of the magnet did not impact the response of the robots,

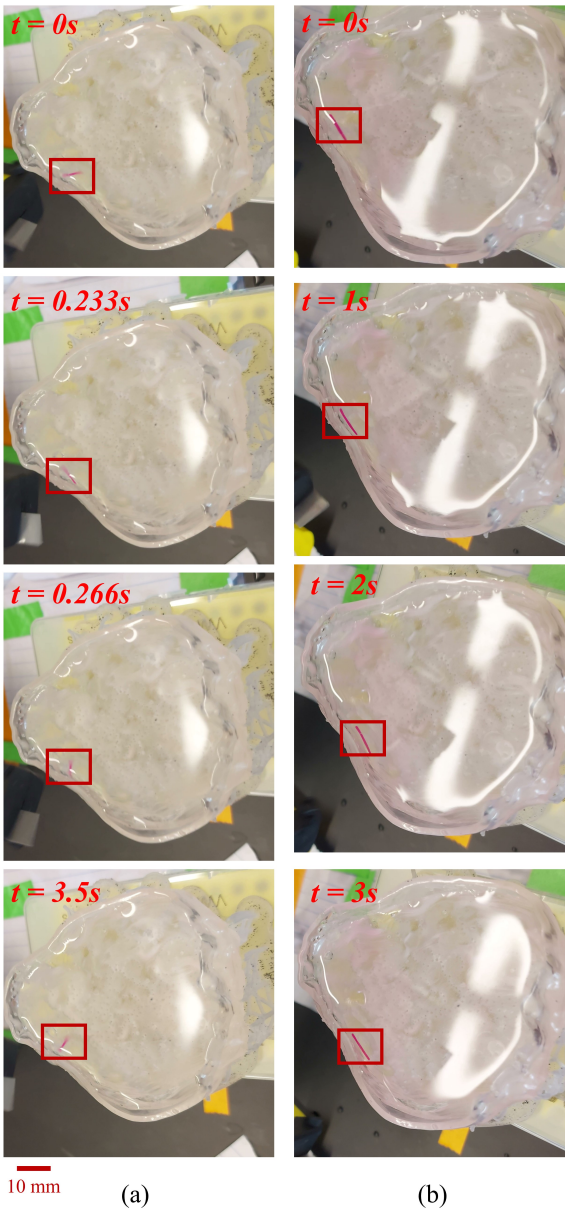


Figure 4.9: Guidance of the fin-like (a) and the screw-like (b) robots in the bladder at 8Hz, the red box indicates the location of the robot

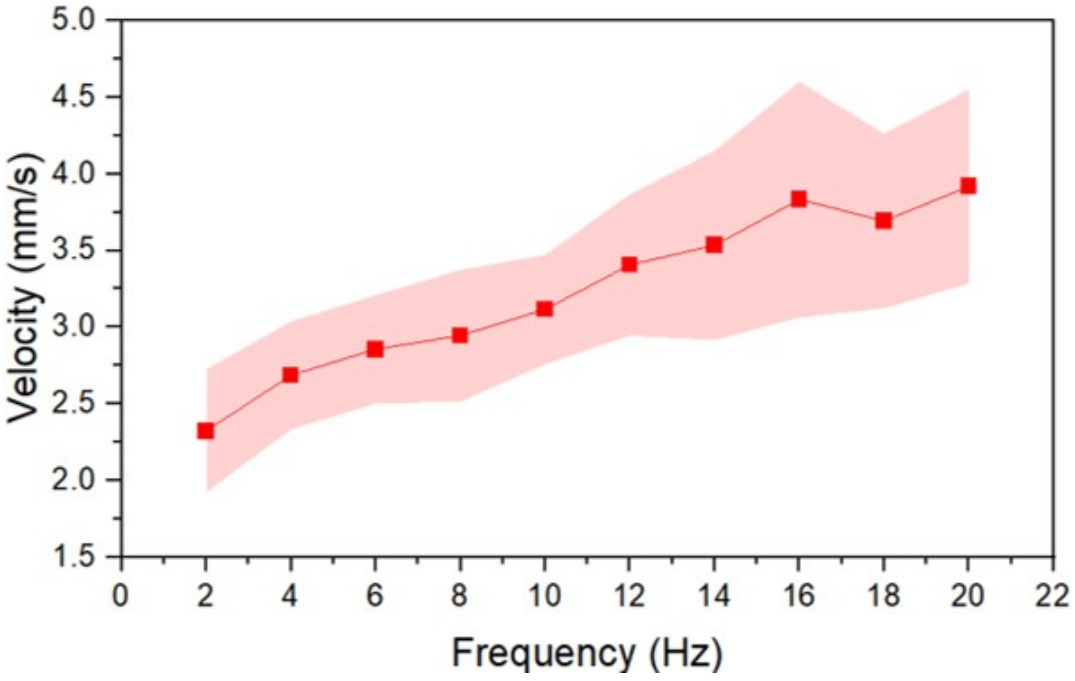


Figure 4.10: Frequency response of the screw-like robots in the bladder

only case 2 was tested where the magnet is embedded between the midpoint and head of the robot. Fig. 4.9 shows the response of the filaments in the bladder where there is no confinement. Tests showed that the screw-like robots move at the same rate as in tubular environments with each actuation frequency, whereas the fin-like robots produce a somersaulting motion that is not very controlled and unpredictable. Furthermore, 5 robots were tested with 3 samples for each robot to compare the velocity of the filaments with the tube. Fig. 4.10 shows that the screw-robots response to actuation in the bladder is similar to their response in a tubular environment.

It should be noted that the robot arm used for these experiments has only 4 degrees of freedom, meaning that it can only produce motion in x, y and z direction and rotate about x, which is not enough to produce a stable and well-defined circular motion around the bladder. Therefore, the robots were tested only along one wall where control was possible. This fact also impacts the experiments which were conducted in the renal pelvis.

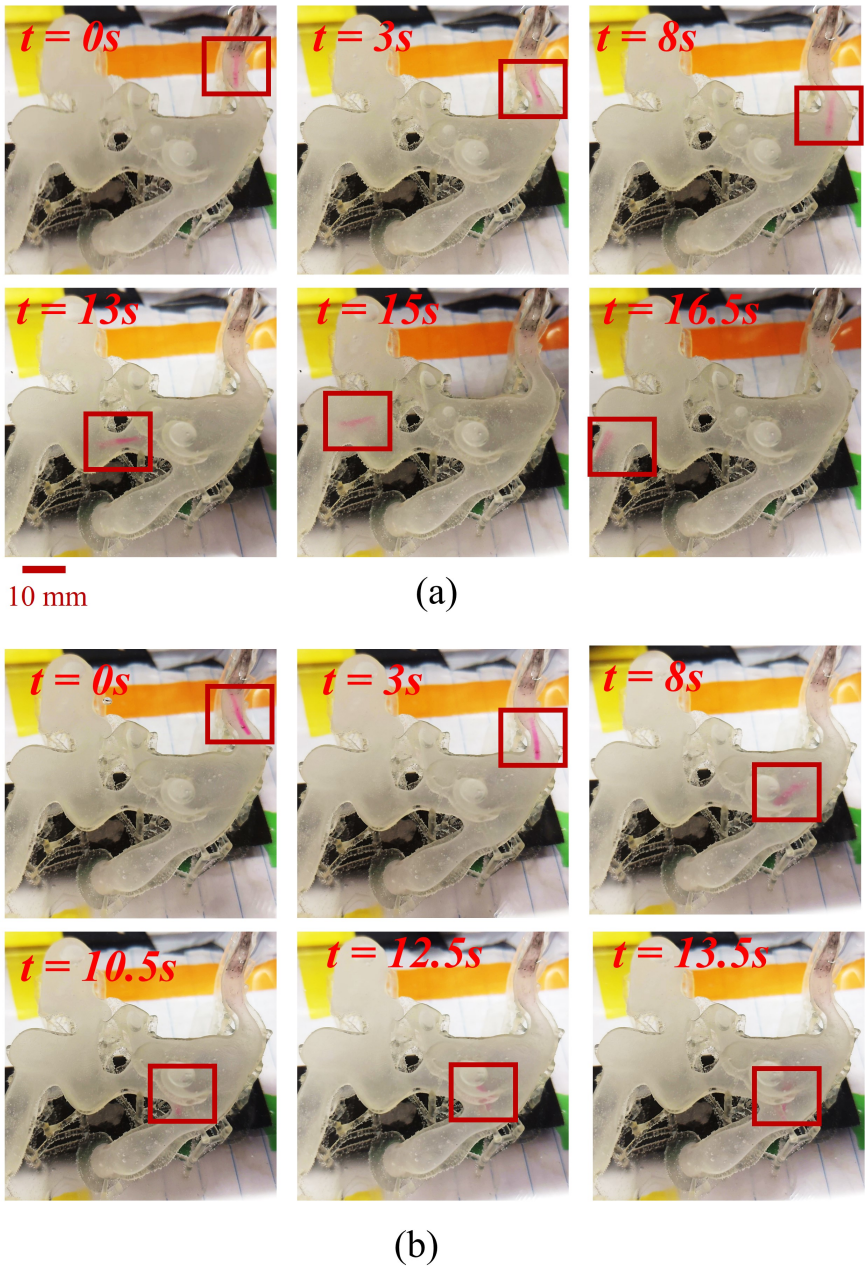


Figure 4.11: Guidance of the screw-like (a) and the fin-like (b) robots in the renal pelvis at 8Hz, the red box indicates the location of the robot

The renal pelvis is a complicated environment where the direction of the actuation has to change and adapt to guide the robots properly. This is only possible with a 6 degree of freedom actuation system. Therefore, the positioning of the magnetic field in this case was done by hand. Since there was no longer precise control over the actuation position, the results are only discussed qualitatively and not quantitatively.

Fig. 4.11 shows the response of the screw-like and the fin-like robots in the renal pelvis. Due to the variability in control and the complexity of the environment, no frequency response was performed, however, the following demonstrations show that the robots can be navigated in the renal pelvis. For the screw-like robot, it can be observed that the robot is controllable, and it can be guided from one end of the renal pelvis to the other end quite well. Which confirms the aforementioned results. The results are also cohesive for the fin-like robots where it can be seen from the image series that the robot wiggles slightly faster than the screw-like robot out of the initial tube, however, when faced with the wider section of the renal pelvis where there is no confinement, it completely fails. As soon as the head of the robot experiences a lack of confinement it stops responding well and when it gets free, it exhibits the same motion as seen in the bladder. Furthermore, it cannot be guided back into the confinement.

It should be noted that the distance required to actuate these filaments in the human body means that only pulling the robots with a strong magnet is not a viable option. Our experiments showed that if only pulling was used as the locomotion mechanism, either the actuator magnet needed to be placed extremely close to the workspace, or it needed to be large and strong, with the size defeating the purpose of a compact actuation system. The chosen locomotion strategy means that the filaments experience the motion induced on them by the actuator magnet coupled with the pulling of the actuator magnet and they are able to swim in the required direction.

The findings of this study have significant implications for the design and deployment of soft robotic filaments in practical applications. The high velocity of fin-like robots makes them ideal for tasks requiring rapid transit through tubular structures, such as targeted drug delivery or inspection of narrow pipelines. However, their reliance on confinement and the associated control challenges necessitate further research to develop advanced control algorithms and feedback systems. On the other hand, the screw-like robots, with their

robust and controllable motion, are better suited for applications where precise positioning and bi-directional movement are required. Their performance suggests potential use in medical devices for minimally invasive procedures or in environmental sensors that need to navigate complex terrain. When the system is fully implemented in a clinical setting, the robots will be guided to the desired location and held in place by a magnetic patch for the duration of treatment. After which one strategy might be to have the GelMa disintegrate and the magnet could be flushed out of the urinary tract safely. This mechanism needs to be studied in future research.

4.2 Filament Deformation

On the side of deformation of the filaments, numerical calculations were performed for the magnet placements illustrated in Fig. 4.12 (a) and (c), using the magnetic field and gradient of the actuator magnet as discussed before. These calculations indicate that for fin-like robots, positioning the magnet closer to the filament's center results in reduced deformation. Conversely, for screw-like robots, placing the magnet nearer to the center increases the deformation. It is noteworthy that the deflection in fin-like robots is 1000 times greater than that in screw-like robots. These results are corroborated by the image series shown in Fig. 4.12 (b) and (d). Even though with the instruments at hand it is not possible to measure the deflection of the filaments, it can be seen from the image series obtained that in the case of the screw-like robots, the camera does not capture any deformation and the graph shows that the deflection is in the micron scale. In addition, in the case of the fin-like robots, the deformation is clearly visible in the image series and the graph shows that the deformation is in the millimeter scale. It should be noted that the modulus of elasticity in this research was assumed as 120kPa as measured in [59]. Furthermore, this modeling is only done statically and does not show the dynamic traveling wave in the filaments.

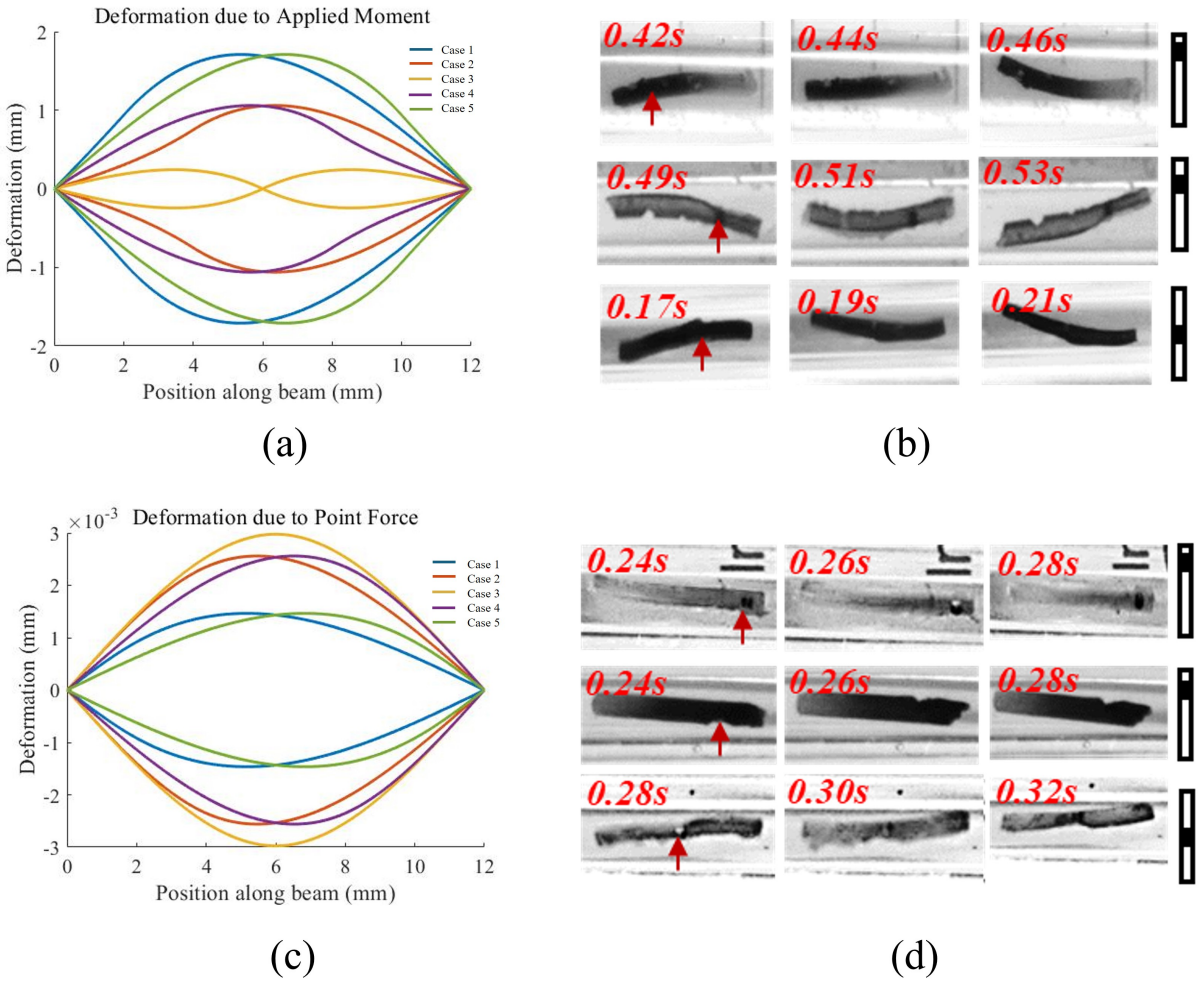


Figure 4.12: (a) Modelling of the deformation of the fin-like robots due to the applied moment (b) Experimental deformation of the fin-like robots due to the applied moment (c) Modelling of the deformation of the screw-like robots due to the applied point force (b) Experimental deformation of the screw-like robots due to the applied point force

4.3 Kidney Stone Dissolution

Initial experiments demonstrated that urease-loaded filaments could increase the pH of synthetic urine from pH 6 to pH 7 within 1 hour of exposure (Fig. 4.13.a). The lowest

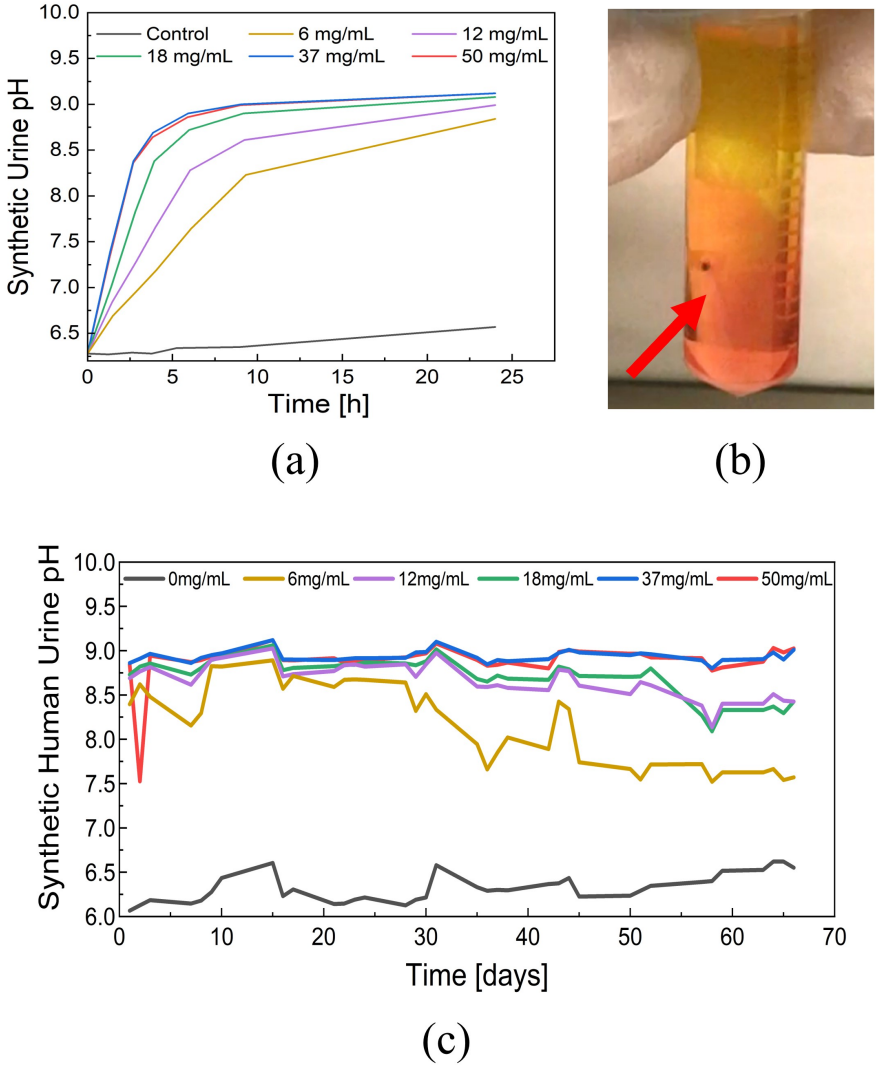


Figure 4.13: (a) Effectiveness of different concentrations of urease embedded in filaments over 24 hours, the increase in urine pH indicates the efficacy of urease (b) Phenol red pH indicator showing the immediate effect of urease in increasing urine pH, arrow shows the filament containing a micromagnet and urease (c) Effectiveness of urease in increasing urine pH in long term, where the filaments were loaded with different concentrations of urease and placed in synthetic urine with pH 6 and the solution was changed every 14 hours

concentration of 6 mg/mL urease was sufficient to generate a drastic pH increase within 24 hours. Figure Fig. 4.13.b shows the immediate pH change of the urine, indicated by the pH indicator phenol red, which changed from yellow to orange in the presence of the enzyme-loaded filament, signaling a shift from acidic to basic pH. Long-term enzymatic activity of the filaments was also tested (Fig. 4.13.c), and it was observed that urease which was loaded in the filaments remained active for over 65 days. The filaments were immersed in synthetic urine at pH 6, and the pH was measured every 24 hours, followed by the replacement of the synthetic urine. This process was repeated for 65 days, demonstrating a consistent pH increase above 7.5 within 24 hours for filaments containing 6-50 mg/mL urease. Control samples containing GelMa filaments without urease showed no pH change, remaining at pH 6.

Following the successful increase in urinary pH, the dissolution of human kidney stones was investigated through in vitro experiments. The experiments included placing the urease-loaded filaments in contact with uric acid stones in 5.5 pH synthetic urine, which is the pH where these stones form. Every 24 hours the pH was measured and the solution was replaced with 5.5 pH synthetic urine. The stone weight reduction was plotted against urine pH to evaluate the correlation between pH and stone reduction.

Figure 4.14.b shows the relationship between pH levels and stone mass reduction percentages across different concentrations of urease per filament, with data points representing concentrations of 0 mg/mL (black squares), 3 mg/mL (red circles), 5 mg/mL (blue triangles), and 8 mg/mL (green inverted triangles). At lower pH values, particularly between 5.5 and 7.0, and with a concentration of 0 mg/mL, stone mass reduction remains relatively low, consistently under 40%, indicating that without urease, the pH has a minimal impact on stone dissolution. As the concentration increases to 3 mg/mL, the effectiveness of urease begins to improve, with stone mass reduction becoming more variable across the pH spectrum. In this case, as pH increases from 6.0 to 7.5, some data points show reductions up to approximately 50%, reflecting a moderate increase in effectiveness at higher pH levels. The 5 mg/mL concentration further amplifies this effect, with a more pronounced increase in stone mass reduction, particularly as the pH approaches and exceeds 7.0. Here, the scatter plot reveals significant reductions in stone mass, with some data points showing reductions as high as 80%, indicating enhanced effectiveness at

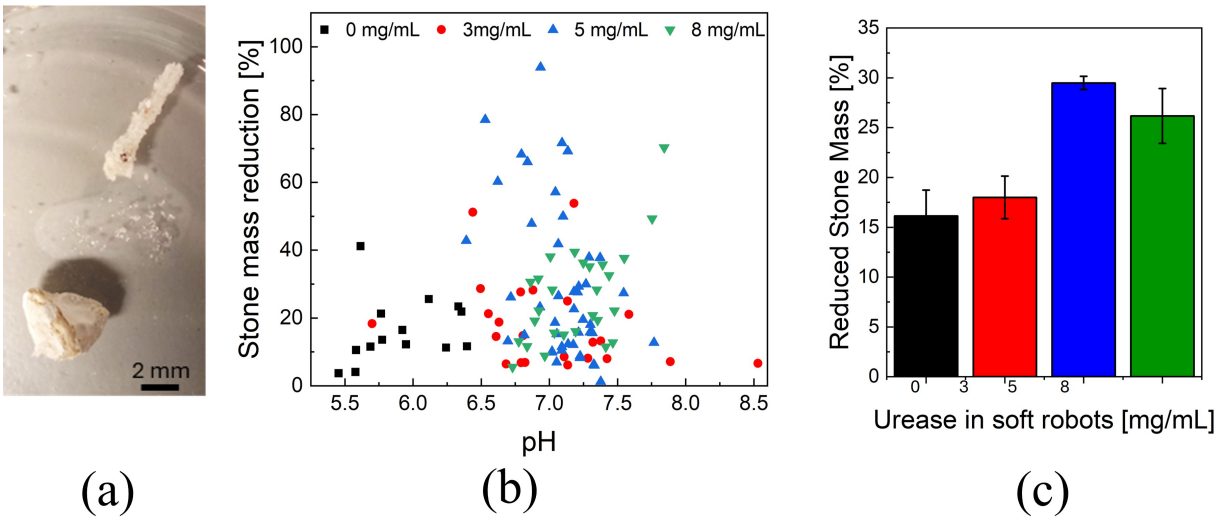


Figure 4.14: (a) Filament and stone after dissolution experiment with crystals formed on the stone (b) Scatter plot showing the most effective dissolution of stones at different pH levels. the percentage of mass reduction for stones in contact with different concentrations of urease is plotted against the average pH after replacement to fresh urine of pH 5.5 every 24h over 4 days (c) Dissolution of kidney stones due to the presence of urease where the average percentage of mass reduction of uric acid stones is compared to different amounts of urease used, showing the amount of 5mg/mL of urease to be the most effective

this concentration, especially in the pH range of 7.0 to 7.5. The highest concentration, 8 mg/mL, shows less stone mass reduction in a higher pH range compared to the previous case. No further stone dissolution was observed when the pH exceeded 7.5 due to high enzymatic activity. This indicates that there is an optimal pH of about 7 where the stone mass reduction occurs and a higher pH does not necessarily mean a more effective stone dissolution.

Optimal stone dissolution was achieved with urease filaments containing 5 mg/mL urease, resulting in an average stone weight reduction of 30% within 5 days and an average urine pH of 7.0 (Fig. 4.14.c). Filaments without urease showed about a 15% reduction in stone mass, likely due to uric acid adsorption onto the GelMa filaments. Filaments with 3

mg/mL urease did not significantly reduce stone mass, despite increasing the urine pH to 6.8. The best outcome was achieved with 5 mg/mL urease, while 8 mg/mL urease resulted in a pH of 7.5 and a 25% reduction in stone mass. These results suggest that maintaining an optimal pH is crucial for stone dissolution, as excessive urease activity and high urine pH can negatively impact stone dissolution. Fig. 4.14.a shows a urease filament and a kidney stone after 5 days of incubation in urine, with recrystallization observed on the filament, possibly indicating dissolved uric acid recrystallizing on the filament. It should be noted that bicarbonate and theobromine were also tested at different concentrations, however, their effect was negligible and very short-term. These chemicals get used up from the GelMa contrary to the urease, which is a catalyst and therefore does not get used up during chemical reactions. This means that within hours, the effect of bicarbonate and theobromine is gone due to the urine flow. Furthermore, citrate is incompatible with GelMa, and bicarbonate makes the filaments brittle and needs to be mixed in great amounts (such as 2 mg/mL) to show any effect. This amount of powder in the GelMa solution makes it very thick which makes it not possible to pipette, and the final shape is also not desirable. Therefore, urease is the better chemical to use for uric acid kidney stone dissolution.

Chapter 5

Conclusions and Future Work

The research presented in this study explored the locomotion capabilities, deformation characteristics, and practical applications of magnetically actuated miniature soft filaments, providing a comprehensive understanding of their behaviors under various conditions and actuation frequencies.

In terms of locomotion, the experiments revealed clear differences between the two configurations when actuated within a tubular environment. Screw-like robots displayed a stable helical rotary motion, closely following the actuator magnet, which enabled them to maneuver effectively and consistently. Their linear velocity plateaued at frequencies above 10 Hz, suggesting a limit beyond which increasing the actuation frequency does not significantly enhance their speed. This behavior indicates that screw-like robots are highly suitable for applications requiring precise control and bi-directional movement, such as in medical devices for minimally invasive procedures or environmental sensors that navigate complex terrains, and they are suitable for applications in the urinary tract.

On the other hand, fin-like robots achieved significantly higher average velocities, reaching up to 18 mm/s compared to the 6 mm/s of screw-like robots. This higher velocity points to their potential effectiveness for tasks requiring rapid movement through confined spaces, such as targeted drug delivery or inspection of narrow pipelines. However, their motion was less predictable and more challenging to control, particularly at higher frequencies. The interaction with the tube walls that propels the fin-like robots forward also introduces vari-

ability in their movement, necessitating the development of advanced control algorithms and feedback systems to fully harness their speed for practical applications.

Interestingly, variations in magnet placement had negligible impact on the frequency response and average velocity of both types of robots. This finding simplifies future experimental setups and design modifications by highlighting that the primary factors influencing locomotion efficiency are the structural design of the robots and the actuation frequency, rather than the precise positioning of the magnet.

The deformation characteristics of the robotic filaments were also investigated, revealing significant differences between the two designs. Numerical calculations and experimental observations showed that fin-like robots experience reduced deformation when their magnets are positioned closer to the filament's center, while screw-like robots exhibit increased deformation in similar conditions. This distinction, with fin-like robots showing deflections on the millimeter scale and screw-like robots on the micron scale, underscores the structural robustness of the screw-like design. This robustness is critical for applications where maintaining the structural integrity of the robots under operational loads is essential.

Beyond locomotion and structural behavior, the study explored the potential application of these robotic filaments in kidney stone dissolution. Urease-loaded filaments effectively increased the pH of synthetic urine, facilitating the dissolution of uric acid kidney stones. Optimal stone dissolution was achieved with urease concentrations of 5 mg/mL, resulting in significant stone weight reduction within five days. The findings indicate that maintaining an optimal pH range is crucial for effective stone dissolution, as excessive urease activity and high urine pH can negatively impact the process. These results suggest that urease-loaded filaments could be a viable solution for non-invasive treatment of kidney stones, offering a controlled and prolonged release of the enzyme to achieve sustained therapeutic effects. The study also explored the limitations of using other chemicals like bicarbonate and theobromine for stone dissolution for localised drug delivery. These substances were found to have negligible and short-term effects compared to urease, primarily due to their rapid release from the GelMa matrix and subsequent washout by urine flow. Citrate was incompatible with GelMa, and high concentrations of bicarbonate made the filaments brittle and difficult to handle. In contrast, urease remained active over extended periods, providing a sustained therapeutic effect that is more suitable for practical

applications.

Overall, the study's findings have great implications for the design and deployment of soft robotic filaments in practical applications. Screw-like robots, with their robust and controllable motion, are well-suited for precise positioning and bi-directional movement tasks. In contrast, fin-like robots, despite their high velocity, require further research to develop control mechanisms that can harness their speed effectively. The successful application of urease-loaded filaments for kidney stone dissolution further demonstrates the potential of these robotic systems in medical and industrial contexts, paving the way for future advancements in soft robotics technology.

Future work includes

- **Dynamic study of the filaments' motion including the deformation and hydrodynamics** To have a better understanding of the motion of the filaments and improve the design of the system there needs to be an in-depth hydrodynamic study of the filaments' motion within dynamic settings, focusing on the deformation behavior induced by fluid flow and wall constraints. The modulus of elasticity of the hydrogel should be measured, which will then be integrated into fluid mechanics and dynamic modeling frameworks. This will enable the detailed investigation of deformation patterns, including the formation and propagation of traveling waves in fin-like robots as they interact with surrounding boundaries. Such analysis is essential to optimize the design and performance of the filaments under realistic physiological conditions.

- **Development of other applications for the filaments, such as treatments of UTI and chronic kidney disease:** The successful application of urease-loaded filaments for kidney stone dissolution opens the possibility of developing similar filament-based treatments for other medical conditions. For instance, filaments could be designed to release antibiotics or anti-inflammatory agents to treat urinary tract infections (UTIs) directly at the site of infection, providing targeted therapy with potentially fewer side effects. Additionally, the filaments could be engineered to release drugs that help manage chronic kidney disease, such as agents that slow disease progression or alleviate symptoms.

- **Exploration of the filaments' application in other target areas:** Beyond the urinary system, these robotic filaments could be adapted for use in various other parts of the body. For example, they might be used to deliver drugs to specific sites in the gastrointestinal tract, or even serve in non-medical applications like inspecting and repairing pipelines in industrial settings. Exploring these applications would involve customizing the filament design and actuation methods to meet the specific requirements of each target area.

- **Exploration of other hydrogels for different medicine and optimization of the filaments' material:** While GelMa hydrogels were used in the current study, other hydrogel materials might offer different benefits, such as improved biocompatibility, drug release profiles, or mechanical properties. Researching and testing various hydrogels could lead to optimized filaments better suited for specific medical applications. This optimization process would involve evaluating factors like the hydrogels' ability to absorb and release drugs, its stability in biological environments, and their interactions with magnetic fields.

- **Acquirement of a 6 DoF robotic arm for the positioning of the actuation:** The current experiments highlighted the limitations of a 4-degree-of-freedom (DoF) robotic arm in navigating complex environments like the renal pelvis. Acquiring a 6 DoF robotic arm allows for more precise and flexible positioning of the actuation magnet, enabling better control of the filament's movement. This upgrade would enhance the system's ability to perform complex tasks and navigate intricate anatomical structures, thereby broadening the scope of potential applications.

- **Development of a sophisticated control system integrated with a localization method (e.g., ultrasound) to make the system fully operable in clinical settings:** To transition from experimental setups to clinical use, a sophisticated control system is necessary. This system should integrate real-time localization methods, such as ultrasound imaging, to track the position of the robotic filaments inside the body accurately. Combining this localization data with advanced control algorithms would allow for precise navigation and actuation of the filaments, ensuring safe and effective operation in a clinical setting.

- **Evaluation of filament performance in dynamic environments:** Testing the performance and robustness of the filaments in more dynamic and varied physiological conditions, such as within the moving environment of the gastrointestinal tract. This would help to understand the adaptability and reliability of the filaments under different physiological motions and pressures, ensuring their effectiveness in real-world scenarios.

- **Ex-vivo experiments in an explanted urinary tract:** The next step in developing the system is testing the filaments in ex-vivo settings. This provides the opportunity to adjust the control system to properly fit a surgical/clinical environment. Furthermore, testing the filaments in an ex-vivo environment details the interactions of the filaments with a real human organ which could defer from an anatomical model.

- **Investigation of long-term biocompatibility and safety:** Conducting extensive in vivo studies to assess the long-term biocompatibility, biodegradability, and safety of the filaments to ensure they do not cause adverse effects when used in the human body.

- **Study into the motion, response, and control of the fin-like robots:** Conducting an in-depth study focused on the motion dynamics, response to actuation frequencies, and control mechanisms of the fin-like robots. Understanding their interaction with different environments and optimizing their design and improving their locomotion efficiency and control precision, addressing the challenges observed in direction control.

Chapter 6

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Robotically Adjustable Magnetic Navigation System for Medical Magnetic Milli/Microrobots

Author: Wonseo Lee

Publication: IEEE/ASME Transactions on Mechatronics

Publisher: IEEE

Date: 11 March 2024

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Generating Rotating Magnetic Fields With a Single Permanent Magnet for Propulsion of Untethered Magnetic Devices in a Lumen

Author: Arthur W. Mahoney
Publication: IEEE Transactions on Robotics
Publisher: IEEE
Date: April 2014

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Magnetic Actuation for Full Dexterity Microrobotic Control Using Rotating Permanent Magnets

Author: Patrick Ryan

Publication: IEEE Transactions on Robotics

Publisher: IEEE

Date: December 2017

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Magnetic localization and control of helical robots for clearing superficial blood clots

Author: Khalil, Islam S. M.; Adel, Alaa

Publication: APL Bioengineering

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Date: May 20, 2019

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Wireless control of magnetic helical microrobots using a rotating-permanent-magnet manipulator

Conference Proceedings: 2010 IEEE International Conference on Robotics and Automation

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A Magnetically Controlled Wireless Optical Oxygen Sensor for Intraocular Measurements

Author: OlgaÇ Ergeneman
Publication: IEEE Sensors Journal
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Conference Proceedings:
2017 IEEE International Conference on Robotics and Automation (ICRA)

Author: Stefano Scheggi

Publisher: IEEE

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Field Model Identification and Control of a Mobile Electromagnet for Remote Actuation of Soft Robots

Author: Alessandro Riccardi

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APPENDICES

Appendix A

Codes

A.1 Arduino Code for Controlling and Monitoring the Actuation Frequency

This code helps the operator to control the actuation frequency by reading the data from the optical encoder on the DC motor and an open-loop feedback control. The operator can adjust the rotation velocity by entering a number between 1-255 (8 bit resolution)

```
#include <Wire.h>
#include <Adafruit_MotorShield.h>
#include <Adafruit_PWMServoDriver.h>
#include <Encoder.h>

Adafruit_MotorShield AFMS = Adafruit_MotorShield();
Adafruit_DCMotor *myMotor = AFMS.getMotor(3);

const int encoderPinA = 2;
const int encoderPinB = 3;
const int encoderResolution = 2000; // Encoder resolution (
   cycles per revolution)
```

```
Encoder myEncoder(encoderPinA, encoderPinB);

const unsigned long predefinedDuration = 3000;
unsigned long startTime = 0;
bool loopRunning = true;

long firstRawPosition = 0;
long lastRawPosition = 0;

unsigned long encoderResetInterval = predefinedDuration;
unsigned long lastEncoderResetTime = 0;

unsigned long convertingFactor = 60000/predefinedDuration;

unsigned long i = 0;
unsigned long currentspeed = 0;

void setup() {
  Serial.begin(9600);
  AFMS.begin();
  myMotor->setSpeed(currentspeed);

}


void loop() {
  // Run the motor continuously
  //Serial.println(i);
  myMotor->run(FORWARD);

   if (Serial.available())
```

```
 {
    int speed = Serial.parseInt();

    switch(speed){
      case 0:
      break;
      default:
      currentspeed = speed;
      break;
    }
    myMotor->setSpeed(currentspeed);

 }

long rawPosition = myEncoder.read();

unsigned long currentTime = millis();
if (currentTime - lastEncoderResetTime >=
   encoderResetInterval) {
  myMotor->run(RELEASE); // Stop the motor


  float totalRotations = static_cast<float>(rawPosition -
     firstRawPosition) / encoderResolution;
  float RPM = totalRotations * convertingFactor;
  float F = RPM/60;

  Serial.print("RPM: ");
  Serial.print(RPM);
  Serial.print(" , Frequency: ");
  Serial.println(F);
```

```
    firstRawPosition = rawPosition;

    lastEncoderResetTime = currentTime;

    myMotor->run(FORWARD);
    myMotor->setSpeed(currentspeed);

  }
  i = i + 1;
  //Serial.println(i);
}
```

A.2 MATLAB Code for Modelling the Field

```
mu0 = 4*pi*1e-7; %Vs/AM = N/A^2
f = 2;
pi = 3.14;

%% Actuator Magnet properties - center at (0,0,0)

M0 = 4.17e6; % volume magnetization [A/m] (it is for the
   actuator magnet)
a = 0.0127/2; % Radius of Cylinder, Unit m
l = 0.0127/4;  % Semi-Length of Cylinder, Unit m


om = 2*pi*f; %rotation speed in rad/s
```

```
Vy = 0.002; %translational speed along y in m/s

%% Robot Magnet Properties - center at (0,0,0.05)

vm = 1.927e-10; %volume of robot magnet
m0 = [0 ;0 ;4.17e6]; % this one is for the robot TO BE CHANGED
m1 = [0; 4.17e6 ;0]; % this one is for the robot TO BE CHANGED

loc = 0.05; %location of center of robot magnet
m = 0.01; % magnet mass in kg (measure the weight at the kab)

%% Counters

tend = 5; % length of time
zc = 0:0.001:loc; %discretization of the distance between the
   center of magnets

t = 0:0.001:tend; %discretization of time (s)

Nz = size(zc,2);
Nt = size(t,2);


%% Calculation of rotating and translating magnetic field

for j = 1:Nt
   for i=1:Nz

        % calculation of field along z
        term1 = (zc(i)+l/2)/(a^2+(zc(i)+l/2)^2)^0.5;
        term2 = (zc(i)-l/2)/(a^2+(zc(i)-l/2)^2)^0.5;
```

```
        Bc = mu0*(M0*l)*(term1-term2)/(2*l); % Tesla = N/Am
        B(i)= Bc; %Tesla
        Bg(i) = Bc *1e4; %Gauss
        Bry(j, i) = 0;
    end


    %Definition of rotating magnetic field
    Brx(j, :) = B*cos(om*j);
    Brz(j, :) = B*sin(om*j);

end

%% Calculation of gradient


for i = 1:Nz
                  if i == 1
        dz(i) = (B(i+1) - B(i)) /0.001;
    elseif i == Nz
        dz(i) = (B(i) - B(i-1)) /0.001;
    else
        dz(i) = (B(i+1) - B(i-1)) /0.001;
                  end
end


for j = 1:Nt


   for i=1:Nz
```

```
      if i == 1
    dBdx(j, i) = (Brx(j, i+1) - Brx(j, i)) /0.001;
elseif i == Nz
    dBdx(j, i) = (Brx(j, i) - Brx(j, i-1)) /0.001;
else
    dBdx(j, i) = (Brx(j, i+1) - Brx(j, i-1)) /0.001;
      end


               if i == 1
    dBdy(j, i) = (Bry(j, i+1) - Bry(j, i)) /0.001;
elseif i == Nz
    dBdy(j, i) = (Bry(j, i) - Bry(j, i-1)) /0.001;
else
    dBdy(j, i) = (Bry(j, i+1) - Bry(j, i-1)) /0.001;
      end


      if i == 1
    dBdy(j, i) = (Bry(j, i+1) - Bry(j, i)) /0.001;
elseif i == Nz
    dBdy(j, i) = (Bry(j, i) - Bry(j, i-1)) /0.001;
else
    dBdy(j, i) = (Bry(j, i+1) - Bry(j, i-1)) /0.001;
      end


   if i == 1
    dBdz(j, i) = (Brz(j, i+1) - Brz(j, i)) /0.001;
elseif i == Nz
```

```
        dBdz(j, i) = (Brz(j, i) - Brz(j, i-1)) /0.001;
    else
        dBdz(j, i) = (Brz(j, i+1) - Brz(j, i-1)) /0.001;
       end


   end

end


%% Calculation of Force or Torque

initial_dBdx = dBdx(1, :);
initial_dBdy = dBdy(1, :);
initial_dBdz = dBdz(1, :);

initial_Brx = zeros(1,51);
initial_Bry = zeros(1,51);
initial_Brz = B(1, :);

initial_grad =  [initial_dBdx; initial_dBdy; initial_dBdz];
initial_B = [initial_Brx; initial_Bry; initial_Brz];

F(:,:) = vm * (m0.* initial_grad);

for i = 1:Nz

    T(:, i) = (vm * cross(m1, [initial_Brx(i) initial_Bry(i)
       initial_Brz(i)]));
end
```