Magnetic actuation for drug delivery in capsule endoscopy

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Magnetic Actuation for Drug Delivery in Capsule Endoscopy

A thesis submitted for the award of the degree of

Doctor of Philosophy

from

UNIVERSITY OF WOLLONGONG

by

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2016
DECLARATION

I, Fredy Munoz, declare that this thesis, submitted in fulfilment of the requirements for the award of Doctor of Philosophy, in the School of Mechanical, Materials and Mechatronic Engineering, University of Wollongong, Australia, is wholly my own work unless otherwise referenced or acknowledged, and has not been submitted for qualifications at any other university or academic institution.

Fredy Munoz

August 2016
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LIST OF PUBLICATIONS DURING THE PhD COURSE


ABSTRACT

The development of a highly controllable drug delivery system (DDS) for wireless capsule endoscopy (WCE) is an important field of research due to its promising features in therapeutic treatment of diseases in the gastrointestinal (GI) tract and drug absorption studies. Before establishing an effective DDS for WCE, several factors need to be considered to set the minimum requirements for the DDS. Operation conditions in the GI tract as well as pharmaceutical factors play a significant role in determining the requirements. In order to facilitate the effective operation of a DDS in the GI tract, at least two mechanisms should be incorporated into a capsule endoscope (CE): an anchoring mechanism to control the capsule position and a drug release mechanism to control variables such as drug release rate, number of doses and amount of drug released.

The literature review indicates that there is inconclusive solution to the development of an active DDS for WCE, despite substantial research being conducted towards their establishment. The aim of this research is to establish an active drug release mechanism for capsule robots by remotely actuating an embedded drug delivery mechanism inside the capsule with an external magnetic field. In particular, this thesis reports on the design, optimization, fabrication and testing of a magnetomechanical system for DDS in WCE. This system allows the active control of an on-board drug release mechanism embedded inside a prototype of capsule robot that would operate in the small intestine of the GI tract. A magnetic linkage is created between the external magnetic system that is located outside the patient’s body and the small permanent magnet that is placed within the capsule robot. This small permanent magnet drives a slider-crank mechanism that is also embedded inside the capsule robot. Therefore, by controlling the relative position and orientation of the external magnetic system with respect to the capsule robot, it is possible to accurately actuate the drug release mechanism. This magnetomechanical system allows the creation of different drug profiles by controlling the release rate, release amount and number of doses which are critical variables to be controlled in an on-demand DDS for WCE.
This thesis presents a novel magnetomechanical system in which the drug release mechanism is driven by magnetic torques. The magnetic linkage between the external magnetic system and the driven magnet has been optimized. The external magnetic system has been optimized in terms of its design, shape, angular positions and dimensions. The shape and dimensions of the driven magnet have also been optimized. All these optimization processes have been carried out using analytical models, which have been validated with numerical solutions and experimental results.

Based on the optimized magnetic linkage, we have fabricated a scale down prototype of the external magnetic system and a prototype of the capsule robot with an on-board drug release mechanism. The experimental results from the proposed magnetomechanical system show that a torque-driven DDS for WCE is feasible and can be used in clinical applications. We tested that the magnetic torque would not be affected if the external magnetic system was scaled up. The optimized magnetic linkage allows further miniaturization of the driven magnet, which not only allows larger operating distances between the external magnetic system and the capsule robot, but also minimizes the weight and volume of the external magnetic system and provides a higher volumetric power density to be transmitted to the driven magnet. Furthermore, the slider-crank mechanism embedded in the capsule robot is fully controllable and can be remotely activated by changing the position and orientation of the external magnetic system. The findings reported in this thesis indicate that the proposed magnetomechanical system is viable for drug delivery in WCE. The outcomes of this study represent a significant step towards minimally invasive technologies for the treatment of diseases in the GI tract.
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Abbreviations

ANOVA: Analysis of Variance
ASMs: Arc-shaped permanent magnets
CE: Capsule endoscope
DDS: Drug delivery system
EPMs: External permanent magnets
FEM: Finite element method
GI: Gastrointestinal
IPM: Internal permanent magnet
MEMS: Micro-electro-mechanical systems
RF: radio frequency
WCE: wireless capsule endoscopy
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1.1 Wireless capsule endoscope

Existing capsule endoscopes (CEs) are used to diagnose diseases in the gastrointestinal (GI) tract, but they are not yet developed to the point where they are able to perform an accurate therapeutic treatment. A typical CE possesses a camera, a battery and electronic circuits that allow physicians to collect pictures of the GI tract while the CE moves through it under natural peristalsis. These pictures are analyzed by experts who determine the medical condition of the GI tract [1].

The commercial CE was introduced in 2001 and since then, several improvements have been made in its image resolution and external communication capabilities [2]. The US FDA has approved three pill-sized bowel capsules (PillCam SB, EndoCapsule, and MiRo capsule) and one esophageal capsule (PillCam ESO). PillCam Colon is a colonic capsule available in Europe and Japan that was cleared in 2014 by the US FDA [3, 4]. All these commercial capsules are imaging devices used for diagnostic purposes in the medical application known as wireless capsule endoscopy (WCE) and a comparison between the capsules is presented in [5].

However, as the capsule is being driven by the natural peristaltic movements of the GI tract, known as passive locomotion, the CEs still miss abnormalities and lesions in the GI tract due to the lack of position, orientation and speed control over the capsule camera. This lack of control over the camera has significant implications for their effectiveness, given that non-inspected areas may lead to incorrect diagnoses [6, 7].

In order to overcome this problem, several systems have been proposed to actively actuate and control the capsule position and orientation in different sections of the GI tract. Some of these proposed systems consist of legged-like mechanisms such that an endoscopic capsule can resist peristaltic forces in narrow sections of the GI tract such as the intestine [8]. In other studies, endoscopic capsules have been covered with magnetic shields of different shapes that can interact with external magnetic fields. These mechanisms have shown promising results in sections of the GI tract.
where the capsule inspects larger areas and move over steep surfaces like the stomach [9, 10]. These proposed systems have been implemented in prototypes and tested in vitro, but will not be available for clinical use until further in vivo tests are conducted.

Similar to the fabrication of CEs for diagnostic purposes, there has recently been considerable interest in the development of mechanisms that can be incorporated into endoscopic capsules to perform additional procedures including biopsy [7, 11, 12] and therapeutic treatments such as drug delivery [1] and surgical interventions [13]. The addition of these features to current CEs will allow clinicians to wirelessly treat diseases of the GI tract, and minimise the discomfort to the patient through this alternative non-invasive procedure [14].

1.2 Research problem

In recent years, the development of a highly controllable drug delivery system (DDS) that would allow clinicians to release an appropriate amount of a drug at specific sections of the GI tract has become an important field of research. These systems can be used in different applications, including the therapeutic treatment of diseases in the GI tract [6, 15], and drug absorption studies, which represent a cost of millions of dollars to the pharmaceutical industry [16].

The fabrication of a remotely actuated DDS is challenging since the CE must operate in a constrained and delicate environment made of live tissue. The DDS has to be embedded in a swallowable capsule whose dimensions impose a restriction on the size of the DDS. The IntelliCap, the IntelliSite and the Enterion capsule all can release drugs (up to 1mL) [17]. The latter one is the most commonly used and remotely controlled device to investigate regional drug absorption since it is capable of delivering a wide range of dosage forms including solutions, suspensions, particulates, and mini tablets [18]. Despite the progress in DDS for WCE, these capsules lack an anchoring mechanism and the released drug is not fully controllable yet.

In order to overcome these limitations, a variety of mechanisms to release drugs at specific regions in the GI tract have been proposed and incorporated in prototypes of CEs recently. Some studies have reported on the remote actuation of drug release
mechanisms fabricated using MEMS technology [19] while other studies have focused on non-mechanical approaches to develop untethered mechanisms [20, 21]. An example of a MEMS system to target and treat pathologies in the GI tract was recently proposed in [8, 17]. This system aims to anchor the capsule and release a liquid drug through a needle.

Most of the MEMS-based systems for DDS in capsule endoscopy incorporate small batteries that are placed inside the capsules to power and actuate the drug release mechanism. Similarly, a small number of studies have also included anchoring systems to allow endoscopic capsule prototypes to firmly attach to the walls of the GI tract before releasing the drug [8]. However, all these mechanisms require power that cannot be supplied for a reasonable time using existing batteries. Consequently, other researchers have investigated wirelessly powering to actuate the MEMS-based systems [22-24].

In other studies, researchers have developed and tested non-mechanical approaches to release drug loads remotely. These proposed systems differ greatly from the MEMS-based systems in that they do not need batteries or wireless power transmission to operate since their actuation relies mostly on chemical interactions that are triggered in response to certain conditions of the environment such as the temperature and pH [21]. Despite the advantage of low power consumption offered by these non-mechanical systems, it remains difficult to control variables such as a release rate, target location, number of doses and exact amount of drug released, which play an important role in on-demand DDS [25, 26].

Endowing CEs with such mechanisms will facilitate the treatment of diseases in the GI tract that are currently not possible with existing tethered endoscopy, bringing benefits to patients and medical practitioners. The patients will be subjected to less discomfort and lower chances for possible side effects and medical practitioners will have a greater control over drug administration, allowing the speed up of treatments and procedures. These mechanisms can also be implemented and adapted in different procedures such as biopsy and therapeutic treatments. Furthermore, a DDS for WCE will offer great benefits in pharmaceutical studies where drug absorption evaluation is
a fundamental part in the creation of new medication [15, 18, 26, 27]. DDS for capsule endoscopy will take the current technology from merely passive diagnostic systems to an active system to perform pharmaceutical procedures [1, 7, 11-13].

Although significant efforts have been made to add features to CEs that would enable physicians to perform diagnostic routines and therapeutic treatments, a range of technical problems still remain unsolved. Specifically, in the development of a highly controllable DDS for WCE, at least two main problems have to be addressed. The first is the implementation of an anchoring mechanism that allows further control over capsule position. The second challenge is to implement a reliable and accurate drug release mechanism whose performance can be fully controlled. Its performance could be measured in terms of the ability to control variables such as release rate, number of doses and amount of drug released [26, 28, 29]. The research problem addressed in this thesis is the establishment of a highly controllable drug release mechanism to be embedded in a CE, and modelling, analysis and design optimisation methodologies needed for this purpose.

1.3. Principal contributions

Within the scope of this thesis, we have proposed, designed, optimized, fabricated and tested a magnetomechanical system that uses magnetic actuation to remotely control a drug release mechanism embedded in a capsule robot. The original contributions of this work are as follows:

1. A novel magnetomechanical system that includes an external magnetic system made of permanent magnets, and a mechanical mechanism that is articulated with an on-board permanent magnet which are embedded inside a prototype of a capsule robot.

2. Different methods were used to optimize the magnetic linkage between the external magnetic system and the driven magnet that is placed within the capsule robot. These methods include the optimization of the design, shape, angular positions and dimensions of the permanent magnets of the external magnetic system and also the shape and size optimization of the driven magnet embedded in the capsule robot.
3. An analysis of the magnetic interactions between the external magnetic system and the driven magnet (i.e., internal permanent magnet (IPM)) are carried out by using analytical models which are effective in conducting parametric studies. This analysis, that also includes a statistical analysis and numerical results, helps to establish guidelines for the establishment of effective magnetic systems that can be used in the propulsion of magnetic devices for medical applications.

4. A number of scaled-down prototypes of the external magnetic system were fabricated to validate the analytical and numerical results of the magnetic interactions. Several prototypes of capsule robots with drug release mechanisms were tested under the external magnetic systems. With these tests, we assessed the capability and feasibility of our proposed system and also found its limitations and its range of operation.

1.4 Thesis outline

Chapter 2, which reviews a number of drug delivery systems proposed in the literature for capsule endoscopy, provides the most recent progress in this field in order to identify the design and functional requirements for the DDS.

Chapter 3 presents the proposed magnetomechanical system to release drug from a capsule robot that would navigate through the GI tract. It also describes the design and optimization of the magnetomechanical system using theoretical and experimental results. We focus on optimizing the magnetic link between the external magnetic system and the IPM. This analysis provides a useful guide for the optimization of feasible magnetic structures.

Chapter 4 presents an analysis of the shapes of the permanent magnets used in the external magnetic system for the purpose of enhancing the magnetic field and subsequently the torque imparted to the IPM. Both theoretical and experimental results are employed to establish the appropriate magnetic system to be used in the remote actuation of a drug release mechanism.

Chapter 5 reports on the optimization methodology used to enhance the magnetic field and torque on the IPM. Specifically, the optimal angular positions and optimal
dimensions are found for the arc-shaped permanent magnets (ASMs) that make up the external magnetic system. The theoretical and experimental results confirm the feasibility of the magnetomechanical system to actively control the drug release mechanism.

Chapter 6 presents the analysis of the magnetic torque on an IPM that is subjected to an optimized external magnetic system that is described in Chapter 5. The first part of this chapter is dedicated to assessing the magnetic torque on an IPM located at any position. This analysis is presented with the assumption that the IPM is not tilted. The second part of the chapter introduces the analysis of the magnetic torque on a tilted IPM, the centre of which can only move in a restricted region of operation. These analyses are important to determine the limitations of the system.

Chapter 7 presents a full analysis of the magnetic torque on an IPM that can have arbitrary position and orientation within the entire region of operation. Analytical models for the rotating magnetic field and the magnetic torque are derived and validated with experimental results. The testing of different prototypes of capsule robots with on-board drug release mechanisms are conducted under different environments. These analyses provide a deep understanding of the functionality of the drug release mechanism.

Chapter 8 presents conclusions drawn from this study and provides recommendations for future research.
Chapter 2
Literature Review

The purpose of this chapter is to review recent research into the development of a DDS for capsule endoscopy and provide a comprehensive comparison among all the different approaches, highlighting their advantages, disadvantages and conclusions. In order to better understand the restrictions and requirements for DDS in capsule endoscopy, and to compare the performance of prototypes of DDS for CE that have been proposed in the literature, a detailed description of the environment under which the DDS would operate along with technical requirements is set out in the following section.

2.1 Operational environment and requirements for DDS

The GI tract can be divided into four sections, the esophagus, stomach, small intestine and large intestine or colon [30] as shown in Fig 2.1. The small intestine possesses three main sub-compartments, that is, the duodenum, jejunum and ileum [18]. Similarly, the large intestine consists of four sections, the ascending colon, transverse colon, descending colon, sigmoid colon and the rectum [27].

The small intestine is about 6 m long, its diameter is 2.5 to 3 cm and the transit time through it is typically 3 hours [18, 31]. The duodenum has a C shaped, and is 30 cm long. Due to its direct connection with the stomach, it is physically more stable than the jejunum and ileum, which are sections that can freely move.
Figure 2.1 Architecture of the GI tract [30].

The jejunum is 2.4 m in length and the ileum is 3.6 m in length and their surface areas are 180 m² and 280 m² respectively [8, 18]. On the other hand, the large intestine is 1.5 m long, its diameter is between 6.3 and 6.5 cm, the transit time though this section is 20 hours and has a reduced surface of approximately 150 m² [18, 27, 30].

The higher surface area of the small intestine enhances its capacity for drug absorption. Thus, this section of the GI tract is of great interest in drug absorption studies that aim to understand the pharmacological behaviours of the majority of molecules administered orally [18]. However, the complex geometry of this section of the GI tract makes it more difficult for conventional endoscopes to pass through it [30]. On the contrary, the large intestine possesses a reduced surface area and lower motility that enhance the mucoadhesion which is a desirable feature to be considered in the development of anchoring systems in capsule endoscopy [32].

Due to the disparities between the sections of the GI tract, different capsule endoscopes have been implemented to target individual sections of the GI tract. Specifically, there are commercial endoscopic capsules that aim to target the esophagus, the small intestine and the colon for medical diagnoses [30]. Since the physical dimensions of the GI tract such as length, diameter and shape, vary throughout the digestive system, capsule endoscopes are restricted in size. This constraint is mainly imposed by the smallest diameter in the GI tract. Existing capsule endoscopes are 11 mm in diameter, 26 mm long, with a volume of 3.0 cm³ and any device with similar dimensions can be considered swallowable [8, 33].

In addition to the size constraints, the transit times in the GI tract vary greatly from one section to another. In order to actively control the transit time of a capsule endoscope, an anchoring or stopping system must be developed and incorporated to wireless capsule endoscopy. Different efforts have been made to allow clinicians to control the position of a capsule endoscope and explore in more details the other areas of interest for a prolonged time. These stopping systems have been developed to meet environmental conditions of each section of the GI tract. For instance, an anchoring system was proposed in [34] for esophageal inspection, a stopping mechanism for
stomach inspection has been presented in [9], while other studies have focused more on the intestine sections [6, 8].

In order to design and develop an accurate drug delivery system for capsule endoscopy, pharmaceutical properties of the administered drug and physiological factors of the GI tract must be considered. Pharmaceutical factors such as dosage form (e.g., liquid or powder compounds) and physiological factors such as gastric emptying rate, fluid, and motility are common factors that affect drug absorption [35].

Changes in the GI tract such as the diameter of the intestine, the pH level, the motility, peristalsis and transit time can occur for several reasons including disease conditions and the aging factor. For instance, gastroesophageal reflux disease is characterized by diminished peristalsis and chronic primary constipation may be associated with reduced intestinal transit rates in the large intestine. In addition to disease conditions, there are also normal changes in the GI tract as the age advances [36].

For instance, a DDS with a passive release mechanism is highly dependent on the fluid availability of the region where the drug is administered. This can be problematic for regions with low fluid like the colon [9]. Thus, a full control over parameters such as timing, duration, release rate, volume of the drug reservoir, number of doses, dosage form and targeted location in a DDS is highly desirable to minimize the dependency on both pharmaceutical and physiological factors [25, 26, 28].

The physiological, mechanical and chemical characteristics of each specific section of the GI tract along with pharmaceutical factors determine the requirements to be met by DDS in capsule endoscopy. Furthermore, clinical motivations such as the need to increase the residence time in the stomach for many therapeutic agents also indicate that active DDS for capsule endoscopes will be of great benefit to patients [37]. Since different DDS have been developed to allow endoscopic capsules to deliver drug at targeted sections of the GI tract, their technical features differ from one to another. In order to compare these proposed DDS in capsule endoscopy, the following variables can be used to measure their performance: release rate, release amount, number of doses, dosage form and if the drug is released in a specific position or over a section.
The controllability of all these variables will offer great advantages in DDS of capsule endoscopes and are discussed in the following section.

2.2 Comparison of existing DDS

The development of an effective DDS for capsule endoscopy should include at least an anchoring mechanism and a release mechanism. The first mechanism would enhance the capsule’s capability to resist peristaltic forces, thus allowing the clinician to actively control the capsule position and orientation at any time. This is a requirement to fully control the transit time which is an environmental factor that varies across the GI tract. The second mechanism would allow the clinician to deliver specific amounts of drug at a target location, thus improving therapeutic effectiveness while minimizing side effects.

Since a number of researchers have focused on the development of anchoring systems and others on the release mechanism, only few have been able to implement prototypes of both mechanisms in a capsule endoscope. If one is to use the frame suggested in [20] for micropump classification, all these systems could be classified into two categories: mechanical and non-mechanical systems. Mechanical systems usually consist of moving parts embedded in the system that include a physical actuator. On the other hand, non-mechanical systems refer to mechanisms that do not require embedded moving parts in the capsule robot to accomplish its design purpose. The following sub-sections will review these categories in more detail for each mechanism.

2.2.1 Mechanical systems for anchoring mechanisms

In [38], the authors proposed a capsule with two legs in the front and two legs in the rear of the capsule body to enhance its steerability as shown in Fig. 2.2 (a). These sets of legs could be deployed to actively control the position of the capsule at any section of the GI tract. A detailed analysis of the leg shapes was included in this study to determine the best possible configuration of the legs around the capsule body. This analysis aimed to develop the less invasive system that would produce the minimal discomfort to the patient. It was found that a leg with a C shaped tip would be the most
appropriate strategy to actively control the position of the capsule. The legged mechanism was designed to reach 40 mm when the legs were completely expanded.

Despite the promising results achieved in this study, several challenging issues were reported. For example, these legs were powered by an on-board battery that actuated a micromotor. Therefore, power consumption and space available within the capsule to house all the electronic components are the main technical drawbacks. In addition, a failure in the synchronization of the legs may cause injury to the GI tract wall since the legs could fold the tissue if they are not controlled correctly. Although this legged mechanism would be adequate to propel a capsule endoscope through any section of the GI tract, due to the legs' length limitation, it would not be suitable as an anchoring mechanism for sections of the GI tract where the average diameter is larger than 40 mm such as the stomach.

Figure 2.2 Legged mechanisms, (a) four legged design [38], (b) twelve legged design [33] and (c) two legged rounded shaped mechanism [8].
In order to reduce the possible damage to the GI tract tissue caused by the legs, it was proposed [33] to increase the number of legs in the capsule endoscope. A twelve legged-mechanism, as shown in Fig. 2.2 (b), was designed, implemented and tested and it was found that this mechanism not only improved the propulsion but also reduced the negative impact that one single leg could cause to the GI tract tissue. This study also suggests that when legs are opened to a diameter of approximately 30–35 mm, the capsule is able to engage the colon wall without damaging it. The improvement in the minimal damage to the tissue is obtained through an increase in the number of legs. This approach required the incorporation of two micro motors that were able to actuate the sets of legs independently. Consequently, the power supply and miniaturization to embed all these electronic parts in the capsule still remain among the major challenges. Furthermore, the possible damage that those legs could cause to the tissue and the legs’ length constraints need to be considered in future studies.

In another attempt to eliminate or at least minimize the legs’ length limitation present in previous studies, [8, 39] developed a mechanical anchoring system that consisted of a two legged rounded shaped mechanism that could be extended from approx. 60 mm and as far as 71.25 mm. These legs when fully deployed, as shown in Fig. 2.2 (c), possess six points of contact with the intestine wall. The rounded ends of the legs aim to minimize the damage to the GI tract tissue. This approach is similar to the anchoring system proposed in [7] where the capsule deploys a four legged-like mechanism but each leg possesses a wider contact area to treat the tissue softer. Although increasing the legs’ length up to 71.25 mm seems to be useful as an anchoring mechanism for more sections of the GI tract, its applicability in the stomach whose average diameter is larger than 71.25 mm would not be possible. Since these legs are powered by a battery and actuated by a micromotor, the miniaturization and power supply are still challenging and focus of further investigation.

As it can be seen, the mechanical anchoring systems, whose working principle consists of deploying legs to allow the capsule to firmly attach to the wall of the GI tract, are powered by an on-board battery that actuates micromotors embedded in the capsule. From the technical perspective, it is challenging to incorporate all these
electronic components in a capsule volume. In order to overcome this issue, [40] presents an analysis and optimization of the electronics required to drive the micromotor. This study reported an area reduction of about 90% in volume for a battery and micromotors so that more actuators and mechanisms can be embedded in the endoscopic capsule to enhance its capabilities.

2.2.2 Non-mechanical systems for anchoring mechanisms

These types of anchoring mechanisms usually exploit magnetic forces between a permanent magnet located inside the capsule and an external magnetic field that could be generated by either an electromagnet or a permanent magnet [41]. For example, in [10] a cylindrical permanent magnet, 10 mm in diameter and 6mm in length was placed in a capsule prototype. An attractive magnetic force was exerted on this magnet by an external permanent magnet that was located as far as 100 mm. It was reported that the capsule could successfully anchor on the surface of a stomach prototype. This proposed anchoring system could be used in any section of the GI tract since the robot capsule shape is reconfigurable [42].

An attempt to use less volume in the capsule and increase the operation distance, a capsule that has incorporated a ring-shaped permanent magnet has been proposed [9]. This hollow magnet gives more space for additional elements to be incorporated in the capsule. The operation distance between the external permanent magnet and the magnet placed in the capsule was 120 mm. This capsule prototype demonstrated the feasibility of an effective stopping system in the stomach but could also work in any other section of the GI tract.

The advantage of these non-mechanical anchoring mechanisms is the simplicity in their implementation since they do not require moving parts to be embedded in the capsule. Consequently, these systems are less susceptible to faults and use less volume in the capsule. On the other hand, its disadvantages include requiring alignment between magnets and the dependency on the operation distance since the magnetic force is drastically affected by the separation between the magnets [43-45].

Finally, some other studies have proposed anchoring mechanisms consisting of a combination of mechanical and non-mechanical systems. For example, [6] and [46]
presented prototypes of endoscopic capsules with legs that were covered with micropillar adhesives coated with silicone oil layer. The addition of such adhesives to the legs improved the ability of the capsule to resist peristaltic forces in the intestine. A similar concept was used in [47] to release a bioadhesive patch to enhance the anchoring capability of the robot capsule.

2.2.3 Mechanical systems for drug release mechanisms

Passive and active DDS are commonly proposed. A passive DDS refers to a mechanism that relies on the environmental conditions present at the target location to discharge the drug reservoir. On the other hand, an active delivery system refers to the ability of the capsule to expel the drug out of the reservoir once its release mechanism is remotely activated. This eliminates the dependency on the diffusion rate of the drug in the environment.

2.2.3.1 Passive mechanical release mechanism

In [48], it was reported that a DDS consisted of two slotted sleeves, the inner and the outer sleeves. A radio frequency (RF) signal generated from a distance of 10 cm activated a resistor that heated a mechanism that allowed the rotation of the inner sleeve when a temperature of 40 °C was reached. When the inner sleeve rotated, its slots aligned with those of the outer sleeve, exposing the drug contained in the inner sleeve (approx. 0.8 mL) to the GI fluid, as shown in Fig. 2.3.

The total volume of this capsule is approximately 2.75 mL since the capsule is 10 mm wide and 35 mm long. Therefore, the ratio Rdc of the volume of drug reservoir to the total volume of the capsule is 0.29. This means that almost 71% of the total capsule volume is used to incorporate the battery, antenna, and electronic components while only 29% is used to load the drug. Some difficulties observed in this study were leakages before the DDS was activated and a slower diffusion rate in the colon probably due to the lack of fluidity and the diminished gut motility in this segment of the GI tract.
An improved DDS was fabricated in [32] to eliminate leakages. This new device consists of two main parts; the outer sleeve which has a hollow plastic cylindrical body and a removable inner cage that fits in the outer sleeve. Almost 70% of the inner cage surface is opened and liquid or powder diffusion can occur through these slots. The inner cage is spring loaded and held in compression with two shape memory alloy wire clips. Activation of the DDS is initiated by placing the capsule onto the remote antenna for 2 min. This signal deforms the wire clips and activates the spring which propels the inner cage out and away from the capsule body and drug can be dispersed from the opened sides and bottom of the cage, as shown in Fig. 2.4.

The released volume of 1 mL contained in a capsule with a total volume of 2.75 mL and the operating distance was 19 cm. The Rdc for this device is 0.36, which represents an improvement with respect to the previous device. However, one of the major drawbacks of this system was the retention of powder drug in the inner cage.
In the previous two passive release mechanisms, the capsules possessed on-board batteries and electronics systems to remotely actuate the DDS. However, to minimize possible faults due to all the components integrated in the capsule, [16] proposed a capsule with a total volume of approximately 0.847 mL that was made of two magnetic parts. These two parts were magnetically attracted to each other with enough magnetic force to keep the capsule closed during its travel through the GI tract. Once the capsule reached the target position, an external magnetic field was used to open the capsule and release 0.34 mL of content as presented in Fig. 2.5. Since magnets are part of the capsule body, this device offers more volume for the drug chamber, and its Rdc is 0.4. However, one of the major drawbacks found in this system was the retention of certain forms of drug that could not be completely released at the target. A more recent work on passive release mechanisms used small soft magnets within a prototype of capsule robot that were activated with an electromagnet and successfully released a payload of 0.78 mL [49]. The Rdc reported in this work is 0.26 and the drug release module can potentially be added to a commercial CE.

2.2.3.2 Active mechanical release mechanism

The aim of an active release mechanism is to have a higher control over the drug release rate, thus making the DDS less dependent on the availability of the fluid in the area of interest.
Figure 2.5 Capsule made of two magnets [16].

For this purpose, several studies have focused on different techniques to propel a piston that would push the drug out of its reservoir [50, 51]. For instance, the drug release mechanism reported in [15] has allowed a drug release chamber of 0.51 mL in a capsule whose size is 10.2 mm in diameter and 30.0 mm in length. The release mechanism consists of a stretchable component that is released when a signal triggered a calorific element in the capsule. This signal was generated from a maximum distance of 1 m and allowed the stretchable component to push the piston that expelled the drug out of the reservoir as shown in Fig. 2.6.

Figure 2.6 Schematic diagram of the remote-controlled capsule. (1) front crust; (2) microscale localizer; (3) energy source; (4) receiver circuit unit; (5) microelectromechanical systems driving device; (6) sealed layer; (7) driving linker; (8) piston; (9) reservoir; (10) back crust; (11) outward diffuse switch; (12) inside diffuse switch [15].

The total volume of the capsule is approximately 2.45 mL. Therefore, the ratio $R_{dc}$ of the volume of the drug reservoir to the total volume of the capsule is 0.208. This means that almost 80% of the total volume is used to incorporate the battery, antenna, electronic components and the piston while only 20% is used to load the drug. A few
disadvantages of this device include its poor reproducible release of the drug due to the usage of the stretchable component and the fact that only one dose can be released at a time.

In order to overcome these two drawbacks, [25] proposed the propulsion of the piston by the pressure of hydrogen gas generated by a small gas producing cell as shown in Fig. 2.7. In this study, a high frequency signal induced current in an oscillating circuit embedded in the capsule. This electrical current activated the gas producing cell that consequently moved the piston forward and emptied the drug reservoir. The results suggest that it is possible to activate the capsule on demand after intervals of some hours and get a reproducible release of the drug. The prototype capsule has a length of about 25 mm and a diameter of 8 mm. Its total volume is 1.25 mL and the drug reservoir volume is 0.17 ml. Therefore, the Rdc is approximately 0.14. Although this device offers the advantage of multiple doses, the lack of control over its activation time makes it less attractive for scenarios where an interval of time of several hours between doses is unacceptable. Another disadvantage is that only 16% of the total volume is used to load the drug.

Space limitation within the capsule is a drawback of the previous approaches. To overcome this issue, [52] proposed a micro-thruster to push the piston rapidly. Because it is the built-up gas pressure generated by the micro-thruster and not the spring-like mechanics that acts on the piston, drug reflex is effectively eliminated. Furthermore, the drug reservoir volume is 0.7 mL in a regular capsule of 25 mm long and with 11 mm in diameter (a total volume of approx. 2.3 mL). Therefore, the Rdc is 0.30, which allocates 30% of the total volume to the drug reservoir. Fig. 2.8 shows the internal parts of this device. One of the disadvantages of this proposed system is that only one dose can be released.

The most significant disadvantages of DDS reported in [15, 25, 52] is that there is no anchoring mechanism. There is no guarantee of holding the capsule at a specific location while activating the release mechanism. The capsule will move forward upon the activation of the drug release mechanism and pose a significant safety problem.
Figure 2.7 Schematic diagram of a remote-controlled capsule with a gas producing cell and a high frequency receiver to control drug release, (1) feed opening, (2) drug reservoir, (3) piston, (4) gas producing cell, (5) high frequency receiver with integrated transistor, (6) resistor [25].

Figure 2.8 Micro-thruster release mechanism [52].

The above studies have reported different techniques to push a piston that releases the drug from the capsule reservoir. Some of these approaches are more efficient in optimizing the volume of the capsule. However, none of them possesses an Rdc higher than 0.40 and only one dose can be released in a short period of time for practical purposes. The controllability of the number of doses and release rate were enhanced in [10, 53], through the usage of magnetic interactions between internal on-board permanent magnets (IPM) and an external permanent magnet (EPM) as shown in Fig. 2.9.

When the EPM moves closer to the capsule, the drug chamber is squeezed and a dose of the drug can be released. When the EPM moves away from the capsule, the restoring force allows the capsule to go back to its previous uncompressed state in which no drug is released. This process can be repeated by controlling the relative position between the EPM and the capsule robot until the chamber is fully empty. Once the drug reservoir is empty, the capsule robot could be, for example, extracted
passively by means of the natural peristaltic force. This proposed system has a drug chamber volume of 0.17 mL and the total volume of the capsule is 2.3 mL. Therefore, the Rdc is 0.07. The advantage of this system is its ability to release multiple doses and a better control of the drug release rate. Despite the feasibility of this system, further miniaturization is required to increase the volume of the drug chamber. In addition, this DDS operates at a maximum distance of 100 mm between EPM and IPMs. Thus, a careful alignment between magnets is required to obtain a repeatable drug release mechanism.

Figure 2.9 Capsule with two IPMs and one EPM [10].

A similar magnetic mechanism was employed in [29] to achieve on-demand concentrations of drugs since controlling the number of doses and release rate would decrease fluctuations in plasma concentration and lower the potential for toxicity. In this study, a thin magnetic membrane was constantly deflected by an external magnetic field. Although this DDS was not designed for capsule endoscopy, its working principle may be applicable to endoscopic capsules. More recently, several prototypes of capsule-like devices, which possess IPMs and are remotely activated by electromagnets, have been used to demonstrate the feasibility of highly controllable drug delivery systems for human blood vessels [54-56]. In these studies the release rate, release amount and number of doses can be remotely controlled, and therefore their principles could be adapted to achieve DDS for WCE.
As it was presented, the development of an active drug release mechanism implied different strategies to supply power to the actuators. When batteries were embedded in the capsule, it reduced the effective volume to be used by the drug reservoir. For the purpose of increasing both the drug reservoir volume and the availability of power in the capsule, some studies have proposed magnetic systems [10, 29, 53], but also some other researchers have investigated wireless power transmission systems [57, 58]. Despite the advancements in this area, it still remains difficult to effectively transmit the required power to actuate all the mechanisms in a capsule endoscope. One of the major issues to overcome has to do with the safety of the live tissues since the human body may absorb part of the power transmitted.

### 2.2.4 Non-Mechanical systems for drug release mechanisms

Similar to the sub-classification of mechanical release mechanisms, the non-mechanical drug release mechanisms can be divided into two categories: passive and active drug delivery systems. Passive release mechanisms consist mainly of chemical interactions that are triggered in response to certain conditions of the environment such as the temperature and pH [21]. In these systems, the manipulation of physicochemical property of compounds is performed to increase intestinal concentration of drugs. This strategy has shown promising results for colon targeting as reported in [59]. However, it remains difficult with these systems to control variables such as a release rate, target location, number of doses and exact amount of drug released, since the properties of the GI tract can vary greatly among the patients [60].

In contrast to a passive release mechanism, the active one is characterized by micropump systems where non-mechanical energy such as magneto-hydrodynamic energy is transformed into kinetic energy. This energy transformation process drives the liquid drug out of the reservoir. The advantage of this approach is that it creates a bigger volume for the drug reservoir but its disadvantage is that the motion of the fluid sample depends on the drug’s physicochemical properties [20]. Table 2.1 compares the different studies reviewed in this work.
As seen in Table 2.1, the dosage form in all the studies varies from liquid to powder compounds. In order to make the DDS less dependent on the dosage form, it would be of great benefit to fabricate a robot capsule platform with the capability of releasing a wide variety of drug compounds (e.g. drugs with different solubility, viscosity, in
liquid, solid and/or gas form) by making little or no modifications to such platform. For instance, hemostatic agents in powder form were released in [61, 62] to achieve hemostasis in the GI tract, while in [63] the cargo released consisted of micro-grippers to achieve biopsy in the stomach. This latter study slightly modified the MASCE platform proposed in [10] to release the micro-grippers and to determine the location of it to estimate the 3D geometrical model [64]. Furthermore, a capsule endoscope generated gas to provide insufflation to the intestine. The gas was produced when liquids and powders were mixed inside the capsule [65].

It can also be seen, that only one study incorporated the anchoring and the release mechanisms in a capsule prototype and two studies reported higher controllability over the release rate, amount and number of doses for WCE. Most of the studies have focused on either the anchoring system or on the release mechanism. The reported findings meet specific requirements of a particular region of interest in the GI tract.

### 2.3 Conclusions

The GI tract represents a challenging environment for the development of an effective anchoring and drug release mechanism. In order to successfully implement a wireless DDS for capsule endoscopy, several factors need to be considered. All these factors were discussed in this chapter and included an overview of the physiological and mechanical properties of the GI tract, pharmaceutical requirements such as release rate, amount of drug, dosage form, number of doses and also size constraints imposed on the capsule along with the technical requirements.

An important number of studies have attempted to implement capsule prototypes that were able to anchor in tubular sections and in more opened regions like the stomach. One of the main advantages in legged-like mechanisms is that due to the on-board battery and micro motors housed in the capsule, the anchoring mechanism can be remotely activated at an adequate distance without severely compromising its functionality. On the other hand, the main advantage of a magnetic actuation system is its more straightforward implementation which makes it less susceptible to electronic faults that can be generated by malfunctions of on-board actuators and power exhaustion [42]. Despite the promising results reported in these studies, further
investigation needs to be conducted to miniaturize electronic elements and mechanical parts that are incorporated in the capsule body.

The reviewed studies do not report data regarding the capability of the proposed capsule endoscopes to release different dosage forms. Thus, it is not possible at the moment to complete a more detailed analysis to assess this functionality. Further studies need to be conducted to evaluate the performance of available capsule endoscopes when the dosage form is changed. In addition, materials to fabricate the drug reservoir and the storing modality to preserve the drug effectiveness are not clearly reported in the literature reviewed in this chapter, and therefore it is worthy of consideration in future work.

Among all the studies presented in this chapter, the results shown in Table 2.1 suggest that [10] is one of the most complete studies that incorporated both the anchoring and the release mechanism. In addition, it is one of the two studies that reports the functionality of releasing multiple doses and allows a higher control over the release rate and amount, which are parameters that should be controlled in an accurate on-demand DDS for WCE. Although the study reported in [10] fulfils more closely the requirements for a DDS set in this literature review, its lowest Rdc ratio suggests that additional optimization of the space in the capsule is required to increase the volume of the drug reservoir. Besides the anchoring and drug release mechanisms, a tracking module should be also included to allow clinicians to target specific sections of the GI tract. If additional sensors and electronic components need to be embedded in the capsule endoscope to track its position and orientation, then each on-board module should be optimized to leave an appropriate volume for the drug reservoir.

Furthermore, one of the most challenging issues to overcome with magnetic systems such as the ones used in [10, 54-56] is the complex interaction between an external magnetic system and the magnets located in the capsule. This is especially relevant when the magnets are not properly aligned or are far away from each other. Therefore, systems that are based on magnetic coupling can be more accurate if a tracking module is incorporated into the entire system. However, magnetic actuation is an
attractive approach to be used in the remote control of mechanisms for medical applications and it is therefore the approach adopted in this work.
Chapter 3
The Proposed Magnetomechanical System

3.1 Requirements for a DDS in WCE
To remotely actuate a DDS for wireless capsule endoscopy, it is necessary to incorporate actuators into the CE that respond to a signal generated outside the patient’s body. Since the CE and all its on-board components have to operate in a delicate and complex environment in the GI tract, they have to be carefully designed and fabricated to fulfil specific requirements. These requirements are closely related to the environmental conditions such as the physiology, anatomy and physicochemical composition of the GI tract. In addition, there are also pharmaceutical requirements that need to be considered in the development of the DDS for CE as it is presented in Chapter 2.

The physical dimensions and volume a CE are typically 11 mm in diameter, 26 mm in length with a total volume of 3 cm$^3$ (i.e., 3 mL) and a drug reservoir volume between 0.17 mL and 1 mL, seems to be a reasonable volume capacity for DDS in WCE [8, 33, 66] (see also Table 2.1, column named Drug Reservoir Volume). The anchoring mechanism should resist axial and radial peristaltic forces of 422 mN and 912 mN, respectively [6, 8]. However, this force should not be too large that can cause excessive pain to the patient. Thus, the pain level of 5 kPa [10] should be taken into account when designing an anchoring system. The drug release mechanism should allow to perform a variety of release profiles, thus allowing the control over the release rate, release amount (1 mL), multiple doses, and several dosage forms (liquids and powders). The desired operating distance should be larger than 200 mm (see Table 2.1). These are the minimum technical requirements for a practical DDS for WCE. This thesis focuses on the design and development of an effective drug release mechanism that is presented in Section 3.2 to meet the above requirements.

3.2 Proposed magnetomechanical system
Magnetic systems have been used in different medical applications because they are considered safe for biological tissues and cells (a threshold of 2 T is recommended for
occupational exposure [12]), and can potentially be scaled down to actuate overly miniaturized systems [67-72]. In order to develop an active and fully controllable DDS for WCE, a magnetomechanical system to actuate a drug release mechanism is proposed in this thesis. The entire DDS for WCE based on magnetic actuation is shown in Fig. 3.1. This system consists of three main components: the external magnetic system made of permanent magnets (A) that surrounds the patient, the drug release module (B) embedded in the robotic capsule (C) and three complementary modules (D) integrated in the robotic capsule. The components of the drug release module are: an internal permanent magnet (IPM), a slider crank mechanism that is connected to the IPM, a drug reservoir to store the drug to be released and an orifice through which the drug is expelled. A magnetic link is created between the external magnetic system and the IPM. Specifically, the external magnetic system generates a rotating magnetic field that exerts a magnetic torque on the IPM as the position and orientation of the external magnetic system are controlled from a joystick. This magnetic torque will cause the IPM to rotate and its rotational movement is converted into a translation movement by the slider crank mechanism which pushes the drug out of the drug reservoir.

Figure 3.1 The main components of the proposed drug delivery system for WCE. A: ring-shaped external magnetic system, B: drug release module, C: the robotic capsule, D: complementary modules within the capsule (anchoring mechanism, active locomotion system and localization and orientation.
detection module), E: patient bed, F: clinician, G: joystick, H: Human Capsule Interface. Point P represents the origin of the general coordinate system XYZ, $\theta_{\text{EPM}}$ is taken with respect to the X axis, and $\phi$ is taken with respect to the Z axis.

The anchoring mechanism, localization module [73, 74], and active locomotion mechanism [75, 76], which are part of the three complementary modules (D) integrated in the robotic capsule, may enhance the accuracy of the drug release module. However, the work conducted in this thesis focuses only on the development of the drug release mechanism as stated in Chapter 1.

Figure 3.2 shows the details of the components embedded in the robotic capsule that would allow the release of drug from the drug reservoir. The slider crank mechanism consists of a piston (B) that is linearly moved by two connecting rods (C) that are attached to two rotating disks (D). The rods have holes on both ends. One end of the rod is inserted in a piston slot and its other end is connected to a pin on a disk. One disk is placed at the top of the IPM (A) and another disk is placed at its bottom side. These disks rotate about the crankshaft when the IPM is driven by the magnetic torque $\tau_{z'}$ (see Fig. 3.2). The crankshaft that is connected to the IPM is also inserted in the hole in the IPM holder. The IPM holder is fixed and attached to the internal wall of the robotic capsule (E). The coordinate system $X'Y'Z'$ is located within the capsule robot and coincides with the centre of the IPM. Specifically, the origin of this coordinate system is placed at the centre of the IPM.
Due to these size restrictions, we propose a dedicated robotic capsule for DDS. In other words, this robotic capsule does not include the image guidance module (e.g., battery, camera and communication capabilities) to perform the screening procedures that are currently achieved with commercial endoscopic capsules. Therefore, our proposed robotic capsule only possesses those specific modules that are relevant to successfully achieve drug delivery (i.e., the three modules (D) shown in Fig. 3.1 along with the drug release module (B) shown in Fig. 3.2) and we aim to create a drug reservoir volume of at least 0.5 mL. Since our proposed robotic capsule is not vision guided, the loop would be closed by using the data from the localization and orientation detection module which is part of the three complementary modules shown in Fig. 3.1. Nevertheless, the inclusion of an image guidance module in our proposed robotic capsule would make the drug release procedure more accurate. The necessity to include multiple on-board modules in the robotic capsule again emphasizes the requirement of miniaturizing the active drug release mechanism as much as possible.

### 3.2.1 Clinical procedure

The proposed clinical procedure is as follows. After the patient’s digestive system is screened and anomalies are detected, the patient would undergo a therapeutic procedure that may include the delivery of drugs at target regions within the digestive system. In this case, the patient lies in a bed and swallows a new robotic capsule that includes a drug release module and the three complementary on-board modules shown in Fig. 3.1 D. Then, the doctor drives the robotic capsule to the target area by controlling its position remotely. To do this, the doctor activates the locomotion system embedded in the capsule while the external magnetic system, the position and orientation of which can be controlled by a joystick, is placed at an appropriate distance from the patient where it transmits no torque on the IPM, thus preventing the activation of the drug release module during this phase. The localization and orientation module within the capsule wirelessly transmits the capsule’s position in real time to a human machine interface. Once the capsule reaches the target area, the doctor activates
remotely the anchoring mechanism within the capsule to make sure it is firmly fixed on the intestine’s surface.

Our magnetically actuated DDS would be suitable for integration with complementary modules that are not based on magnetics. If any of the complementary modules relies on magnetics for their operation, then careful design and advanced control strategies are needed to fully control the capsule robot. Therefore, the activation and deactivation of these complementary modules must be compatible with the magnetically actuated DDS. The integration of multiple magnetically compatible modules and functionalities for capsule robots is still an open area for further research. For instance, the integration of a magnetic-based tracking system with active locomotion was recently reported with promising results obtained from prototypes of capsule robots [77, 78]. This clearly suggests that our magnetically actuated DDS may be also suitable for integration with complementary modules that are based on magnetic coupling.

After the robotic capsule is properly anchored, the doctor uses the joystick to place the external magnetic system in the correct position and with the correct orientation to activate the drug release module (i.e., to impart a magnetic torque \( \tau_m \) to the IPM as shown in Fig. 3.2). This activation can be achieved by following the next two sequential steps: first, a coordinate system \( X_aY_aZ_a \) (shown in Fig. 3.3), that is associated with the external magnetic system, is adjusted with respect to the general reference system \( XYZ \) shown in Fig. 3.1. Second, the external magnetic system, which can be powered by motors, starts rotating about its \( Z_a \) axis, generating in this way a rotating magnetic field that can impart a magnetic torque \( \tau_{z'} \) to the IPM. The rotation of the IPM about its axial axis \( Z' \) is converted into a linear movement by the slider crank mechanism and the piston pushes the drug out of the reservoir. By controlling the external magnetic system's rotational speed and direction (clockwise or counter clockwise), the doctor is able to control the release rate, release amount and number of doses. These are highly desirable variables to be controlled in an on-demand DDS for WCE to produce different drug profiles [66]. Finally, after the drug is released, the doctor deactivates the anchoring mechanism and reactivates the locomotion module to propel the robotic capsule.
The details and optimization of the external magnetic system and the IPM are presented in Chapters 3-5. Specifically, we dedicate Section 3.3 to model the magnetic link between permanent magnets and Section 3.4 to model the mechanical system (i.e. the slider crank mechanism), followed by a design optimization of the external magnetic system and the shape optimization of the IPM in Sections 3.5-3.6, respectively. We draw conclusions on these preliminary optimization processes in Section 3.7.

3.3 Modeling the magnetic system

3.3.1 Magnetic actuation

Due to the demanding requirements and the limited volume available in a CE, the least complex drug release system is to be embedded in the CE, and the actuation problem is shifted to the exterior of the patient’s body [79]. As presented in Chapter 2, most drug delivery prototypes have used micro-motors, and electronic devices powered by micro-batteries inside the CE. All these electronic components still occupy a significant volume and are more prone to electrical malfunctions [42]. A simpler actuation system that has shown promising results in different medical applications is based on
exploiting magnetic links [80]. There are several advantages when using magnetic systems, including the free availability of energy, and the harmlessness to the human body [81, 82]. For these advantages and the promising results of them presented in Chapter 2, we use magnetic actuation as the approach in this thesis to remotely control a drug release mechanism in a capsule robot for WCE. The source of the magnetic field and the type of actuation are chosen in Section 3.3.2.

3.3.2 Source of the magnetic field

Magnetic systems conform with the idea of integrating simple components in the CE and shifting the actuation problem to the exterior of the patient’s body [42, 63, 64].

A basic description of magnetic interactions is given by [83]

\[
\tau = \frac{\mathbf{V}(\mathbf{m} \times \mathbf{B})}{\mu_0} \quad [\text{N} \cdot \text{m}]
\]

\[
\mathbf{F} = \nabla \mathbf{V}(\mathbf{m} \cdot \mathbf{B}) / \mu_0 \quad [\text{N}]
\]

where \( \mathbf{F} \) and \( \tau \) are the magnetic force and magnetic torque, respectively. The force and torque vectors are exerted on a permanent magnet of volume \( \mathbf{V} \) and with a magnetization vector \( \mathbf{m} \) with its magnitude given in T. \( \mathbf{B} \) is the magnetic flux density of the external magnetic field and is related to the external magnetic intensity \( \mathbf{H} \)

\[
\mathbf{B} = \mu_0 \mu_r \mathbf{H} \quad [\text{T}]
\]

where \( \mu_0 = 4\pi \times 10^{-7} \, \text{H/m} \) is the free-space permeability, and \( \mu_r \) is the relative permeability of a material. For a permanent magnet, its magnetization \( \mathbf{m} \) is almost constant under normal working conditions, unless it is demagnetized by a strong magnetic field or heated over its Curie temperature.

Equations 3.1-3.2 show that it is possible to control \( \mathbf{F} \) and \( \tau \) by changing \( \mathbf{V} \) and \( \mathbf{m} \) of the permanent magnet that is exposed to an adjustable external magnetic field \( \mathbf{B} \). The torque \( \tau \) will tend to orient the vector \( \mathbf{m} \) along \( \mathbf{B} \) and can generate a rotational movement on the permanent magnet. On the other hand, the force \( \mathbf{F} \) will tend to produce a translational movement on the magnet. Furthermore, these two equations also show that the volume of the permanent magnet \( \mathbf{V} \) can be decreased while at the same time \( \mathbf{B} \) is adjusted to obtain the same force and torque. These changes could allow the miniaturization of components to be placed in a CE while \( \mathbf{B} \) is compensated from the outside of the body accordingly.
The external magnetic field \( \mathbf{B} \) can be generated by electromagnets or permanent magnets. The advantage of electromagnets is that the magnetic field direction and magnitude can easily be and precisely controlled. However, its biggest disadvantage is the poor capability to generate higher magnitudes of magnetic fields. On the other hand, small permanent magnets can generate higher magnitudes of magnetic fields compared with electromagnets. Since \( \mathbf{B} \) decreases rapidly with the distance [43], and an appropriate operating distance is one of the main requirements in DDS for WCE, the generation of magnetic fields by external permanent magnets (EPMs) rather than electromagnets appears to be more suitable for this application.

From Eqs. 3.1-3.2, the manipulation of the magnetic force is more difficult than the controllability of the magnetic torque because the former is based on controlling the field gradient while the latter is based on the control of a rotational magnetic field [79]. Some researchers have proposed the actuation of DDS by magnetic forces [10, 29]. To the best of our knowledge, there are no studies using the magnetic torques for DDS in WCE, although magnetic torques have been used to activate other mechanisms such as a biopsy module for WCE [12] and a DDS for a device that is to target human blood vessels [54-56]. The strategy used in this study consists of exploiting magnetic torques exerted on an IPM when it is placed in a rotating magnetic field generated by an external magnetic system that consists of permanent magnets.

Permanent magnets have different shapes, dimensions, and magnetization grades. The next two subsections (subsections 3.3.3-3.3.4) describe cuboidal, cylindrical, ring-shaped, arc-shaped permanent magnets and arrays of them which are considered in this thesis for the magnetic interactions between the EPMs and IPMs to actuate the drug release mechanism.

### 3.3.3 Cuboidal and cylindrical external permanent magnets

Several techniques exist to analytically calculate the magnetic field and the magnetic force between two permanent magnets. The most common approaches consist of representing the magnets by equivalent magnetic charges or currents, the principle of virtual work, or the use of Kelvin’s formula [84, 85]. By using the magnetic charge model, [86] reported the magnetic interaction between two permanent magnets. First
of all, the external magnetic field produced by the first permanent magnet of volume \( V_1 \) and magnetization \( \mathbf{m}_1(x') \) is calculated as

\[
\mathbf{B}(x) = \frac{1}{4\pi} \int_{V_1} \frac{\rho_{M_1}(x')(x-x')}{|x-x'|^3} \, dV_1' + \frac{1}{4\pi} \int_{S_1} \frac{\sigma_{M_1}(x')(x-x')}{|x-x'|^3} \, dS_1' \quad [T] \tag{3.4}
\]

where \( x \) is the observation point, \( x' \) is the source point, and the integration is over the volume for which the magnetization exists. \( S_1 \) defines the surface that bounds \( V_1 \) (the surface of the first magnet that generates \( \mathbf{B}(x) \)). \( \rho_{M_i} \) and \( \sigma_{M_i} \) are defined as the volume and surface charge densities respectively, and are given by

\[
\rho_{M_i}(x') = -\nabla' \cdot \mathbf{m}_i(x') \quad [T/m^3] \tag{3.5}
\]

\[
\sigma_{M_i}(x') = \mathbf{m}_i(x') \cdot \mathbf{n} \quad [T/m^2] \tag{3.6}
\]

where \( \mathbf{n} \) is the normalized vector perpendicular to the surface of the magnet, \( i=1 \) or \( 2 \), and \( \nabla' \) operates on the primed coordinates. If a second permanent magnet of volume \( V_2 \) and magnetization \( \mathbf{m}_2(x') \) is exposed to \( \mathbf{B}(x) \), then the magnetic force exerted on the second magnet is given by

\[
\mathbf{F}(x) = \frac{1}{\mu_0} \left[ \int_{V_2} \rho_{M_2}(x)\mathbf{B}(x) \, dV_2 + \int_{S_2} \sigma_{M_2}(x)\mathbf{B}(x) \, dS_2 \right] \quad [N] \tag{3.7}
\]

Eqs. 3.4-3.7 can be used to calculate the magnetic interactions between permanent magnets, and the magnetic torque can be derived directly from the magnetic force. In addition, Eq. 3.4 indicates that the external magnetic field produced by a permanent magnet depends on the geometrical parameters of the magnet (i.e., its shape and dimension which are taken into account as the functions are integrated over the surface \( S \) and volume \( V \)), its magnetization densities and also varies with the distance to the point of observation. Furthermore, Eq. 3.7 indicates that the magnetic force on a second magnet exposed to the external magnetic field is also a function of the same variables (i.e., dependent on geometrical parameters and magnetization densities of the second permanent magnet).

For instance, the analysis of the magnetic interactions (magnetic force and torque) between two cuboidal permanent magnets, as shown in Fig. 3.4, has been carried out by deriving analytical solutions that use the magnetic charge model (also known as the Coulombian model) [87]. Similarly, the analysis of the magnetic force between two cylindrical permanent magnets, as shown in Fig. 3.5, has been conducted by deriving analytical solutions that use the magnetic current model (also known as the Amperian
model) and the Biot-Savart’s law [84]. Although these studies are conducted with the assumption that the permanent magnets are not rotated but their edges are parallel, they show that magnetic forces and torques depend on the geometrical parameters, magnetization grades and the relative distance between them.

Figure 3.4 Cuboidal permanent magnets parallel magnetized [87].

Figure 3.5 Lateral force $F_l$ and axial force $F_a$ between cylindrical magnets [84].

Furthermore, analytical solutions for magnetic forces between two cuboidal magnets and two cylindrical magnets can be obtained easily. This advantage has been exploited to conduct parametric studies that help to establish design guidelines which can be useful in applications that rely on magnetic forces [84].
Cuboidal and cylindrical magnets are part of basic magnetic structures, and the analytical expressions for magnetic forces and torques in such systems are very helpful to design and optimize their shapes and dimensions. Other more complex magnetic systems consist of arrays of linear and planar magnets (also known as Halbach arrays), as shown in Fig. 3.6, which have been studied to enhance the magnitude of the magnetic field generated in certain regions [88, 89]. These arrays can be analyzed by using the superposition principle—summing the contributions from all of the individual permanent magnets [86, 87].

Figure 3.6 (a) linear array of cuboidal magnets with 90° magnetization rotations, (b) planar array of cuboidal magnets with magnetization directions as the superposition of two orthogonal linear arrays [88].

One advantage of these arrays is that they can focus the magnetic field on one side of the arrangement and reduce or eliminate the magnetic field on the opposite side as shown in Fig. 3.7.

Figure 3.7 Magnetic field lines for a linear array of cubid magnets with 45° magnetization rotations [88]. The magnetic field is focused above the array and is diminished on the reverse side.
These magnet arrays have been used in various applications, including medical applications [80], due to the ability of increasing the magnetic field on one side. These magnetic systems can also be optimized in terms of dimensions of each magnet, number of magnets to be used and the magnetization direction of the magnets. Although the analytical models can be complex, their solution is straightforward. This is of great advantage in the design and optimization of different magnetic systems such as linear and planar arrays [90]. However, when the magnetic system consists of magnets with non-trivial shapes, lack of symmetry or when the system is composed of multiple non-trivial shaped permanent magnets, finding the analytical expression for the magnetic interaction can become tedious. For this reason, often in the design of magnetic systems, numerical solutions such as finite element method (FEM) are used rather than analytical methods. Therefore, in this study, both approaches (analytical solutions and FEM solutions) are used.

The generation of the external magnetic field by means of an array of permanent magnets is a feasible way to actuate a single permanent magnet located inside a CE. This approach of using an array of permanent magnets as the source of the external magnetic field is consistent with the idea of shifting the actuation problem to the exterior of the patient’s body, presented in Section 3.3.1.

3.3.4 Ring-shaped and arc-shaped external permanent magnets

The axial magnetic force $F_z$ between two ring-shaped magnets (axially, radially and perpendicularly magnetized, as shown in Figs. 3.8-3.10, respectively) has been studied previously [91-93]. It has been found, through analytical solutions, that the interaction between ring-shaped magnets with perpendicular magnetization offers a higher axial force $F_z$ than the systems shown in Figs. 3.8-3.9.
Figure 3.8 Ring-shaped magnets axially magnetized [92].

Figure 3.9 Ring-shaped magnets radially magnetized [93].
The previous three magnetic systems are the basis of a more complex magnetic system composed of a stack of those rings as shown in Fig. 3-11. This array of magnets can offer even a higher axial force $F_z$, as reported in [91].

Similarly, magnetic coupling among arrays of arc-shaped permanent magnets (ASMs) with different magnetization directions and different air gaps, as shown in Figs. 3.12-
3.13, has been previously studied to increase magnetic forces and the transmission of high density torques [94]. For instance, it has been found that in cylindrical air gaps, the best configuration to obtain the highest magnetic torque consists of ASMs radially magnetized as shown in Fig. 3.12 (a). These results are obtained by using analytical models of a single ASM, as shown in Fig. 3.14, and the superposition principle [95].

![Figure 3.12 Magnetic coupling systems with cylindrical air gaps; (a) ASMs radially magnetized, (b) ASMs tangentially magnetized , (c) ASMs axially magnetized [94].](image)

![Figure 3.13 Magnetic coupling systems with plane air gaps; a) ASMs radially magnetized, b) ASMs tangentially magnetized , c) ASMs axially magnetized [94].](image)
As presented before, there are a variety of magnetic systems that have been studied and used for different applications [80, 92]. Depending on the specific application and the requirements to be met, one magnetic system or a combination of them could be more suitable than other. For this reason, the analytical expressions of such systems ease the design and optimization process. All these analytical models can be programmed in Matlab and the results can be also compared with numerical simulations from a finite element method software such as COMSOL [12, 84, 88].

In the particular application of a DDS for WCE, a ring-shaped structure made of an array of EPMs seems to be suitable as the source of the external magnetic field due to the promising results in the transmission of high density torques. Therefore, in this thesis, we propose the use of a ring-shaped structure for the external magnetic system as shown in Fig. 3.1. Each permanent magnet of the external magnetic system can have a different shape, dimension, and magnetization grade. In the same way, the IPM can have different geometrical parameters and magnetization grade. Both external magnetic system and IPM can be optimized in terms of their geometrical parameters and positions within the entire magnetic system to meet the minimum requirements for DDS in WCE.
3.4 Modeling the mechanical system

The IPM will be connected to a slider crank mechanism shown in Fig. 3.15. This mechanism is used to transform a rotational movement into a translational movement. Therefore, when the IPM is rotated by the EPMs, the magnetic torque $\tau_{zm}$ imparted to the IPM (shown in Fig. 3.2) is transmitted to the crank-shaft and converted into a piston force that will be used to release the drug from the reservoir.

![Slider crank mechanism](image)

Figure 3.15 Slider crank mechanism [96]. $0 \leq \alpha < 2\pi$ and the mechanism can be actuated in the clockwise or counterclockwise directions.

Designating the point B as the piston position $x$, we find

$$x = R \cos \alpha + L \sqrt{1 - \left(\frac{R \sin \alpha}{L}\right)^2}$$  \hspace{1cm} (3.8)

with

$$\alpha = \omega t$$  \hspace{1cm} (3.9)

where $\alpha$, $\omega$, $R$ and $L$ designate the crank angle, the angular velocity of the crank, the lengths of the crank and the connecting rod, respectively [96]. The centre of the crank is aligned with the $Z'$ axis (i.e., the axial axis) of the IPM shown in Fig. 3.2.

By using the law of cosines, we can also express the crank angle $\alpha$ as a function of $x$ (i.e., the position of point B) as

$$\alpha = \cos^{-1}\left(\frac{R^2 + x^2 - L^2}{2Rx}\right)$$  \hspace{1cm} (3.10)

The piston acceleration can be obtained by differentiating two times Eq. 3.8. If the ratio $R/L$ is small (i.e. $1/3$ or $1/4$), the piston acceleration can be approximated as

$$\ddot{x} = -R\dot{\alpha}(\sin\alpha + \frac{R}{2L}\sin 2\alpha) - R\omega^2(\cos\alpha + \frac{R}{L}\cos 2\alpha)$$  \hspace{1cm} (3.11)

The drug load exerts a force $P$ acting on the piston as shown in Fig. 3.15. In order to
simplify the kinetics of the slider-crank mechanism, we assume that the gravity forces are zero. Designating $m_{lb}$ as the equivalent mass of the rod concentrated at point B and $m_p$ as the piston mass, we find

$$F_d = (m_{lb} + m_p)x\ddot{x}$$ \hspace{1cm} (3.12)

$$\tau_c = (F_d + P)x\tan \phi_c + I\dot{\alpha}$$ \hspace{1cm} (3.13)

where $I$ represents the moment of inertia of the crank and $\tau_c$ represents the crankshaft torque; the counterclockwise direction is positive. Equations 3.8-3.13 describe the kinematics and kinetics of the slider-crank mechanism which can be actuated in both the clockwise and counterclockwise directions.

### 3.5 Design optimization of the external magnetic system

#### 3.5.1 Challenges in magnetic actuation for DDS in WCE

The fabrication of a safe and effective DDS for WCE has received significant attention in the research community and the pharmaceutical industry in the last two decades. WCE is used mainly for diagnostic purposes and has proved to be an effective, non-invasive tool for the evaluation of the small intestine [97]. However, this technology has not been developed to perform therapeutic procedures such as drug delivery to targeted sections of the GI tract. The need to include such an advanced feature in a CE is important to treat a variety of diseases in the GI tract and to conduct drug absorption studies [66]. Therefore, the development of a fully controllable DDS is highly desirable in the next generation of WCE [43-45].

Several approaches have been proposed for the development of DDS in WCE including magnetic systems. As presented in [66], at least two mechanisms should be included in the fabrication of an accurate DDS: an anchoring mechanism to resist the peristaltic forces in the GI tract and a drug release mechanism that delivers different drug profiles. The inclusion of both mechanisms in a CE has been challenging due to several limitations including volume constraints imposed by the capsule size, the delicate environment of operation, the operating distance and the controllability of variables such as the release rate, release amount and number of doses which are essential to produce different drug profiles.
A number of studies have reported promising results on magnetic coupling concepts in medical applications [98]. For instance, in [41], a cylindrical magnet radially magnetized was able to interact at a maximum distance of approximately 15 cm from the magnets’ centres, with two small cylindrical magnets mounted on a prototype of tethered endoscopic capsule for a wireless endoluminal application. A similar work is presented in [99] where a cylindrical magnet radially magnetized interacted, at a maximum distance of 3 cm measured from the magnets’ surfaces, with other three small cylindrical magnets. A large permanent magnet is placed in a device outside the human body and the three small magnets are embedded in a device that is to be inserted into the body.

Recently, a magnetic system consisting of two cubic magnets that actuated two small cylindrical magnets is proposed in [100] for a laparo-endoscopic application. The two small magnets are placed inside of a prototype of endoscope while the cubic magnets are used as the external source of magnetic field and are positioned 3 cm away from the endoscope. In addition to these medical applications, a number of researchers have proposed magnetic systems to actuate endoscopic capsules. In [9], authors developed and tested the magnetic interaction between an external cylindrical magnet and a small cylindrical magnet housed in a CE to control its trajectory. External permanent magnets (EPMs) that interact with four and two small cylindrical magnets are used for biopsy purposes in [12, 63], respectively. Wireless insufflation is achieved when a cylindrical magnet actuated two small magnets located inside a prototype of CE [65].

Despite all these efforts focused on exploiting magnetic linkages between permanent magnets to enhance some desirable capabilities in WCE, very little has been accomplished in regards to magnetic actuation for a drug release mechanism in WCE. An exception is presented in [10], where two small cylindrical magnets can squeeze drug out of a chamber when an external magnet is brought closer. This system can potentially allow clinicians to generate a variety of drug profiles by controlling the release rate, release amount and number of doses. In terms of flexibility to control such important variables in an on-demand DDS, its capability is similar to the endoscopic capsule studied in [51]. However, its actuation system is different.
Although magnetic coupling between permanent magnets has been successfully used to wirelessly actuate a diversity of mechanisms in WCE, some difficulties still remain. For instance, in all the studies [9, 10, 12, 41, 51, 63, 65, 99, 100], the maximum operating distance is 150 mm. This distance might be appropriate for some medical applications, but might not be the case for WCE. Another limitation when mounting permanent magnets inside of a CE is the volume constraint of the capsule which is approximately 3.0 cm³ [66]. Therefore, the miniaturization of the on-board magnets is necessary not only to reduce the volume occupied by the magnets but also to leave enough useful volume for additional components such as the drug reservoir or any other modules.

In order to cope with these two current limitations (the increase of the operating distance and the miniaturization of the IPM), a design optimization of the external magnetic system is conducted in this section (Section 3.5). This optimization aims to increase the magnitude of the external magnetic field so that higher torques can be imparted to a small IPM embedded in the capsule robot.

### 3.5.2 The magnetic system and the operation principle

In the literature, most approaches have used cylindrical magnets as the external magnetic source [9, 10, 12, 41, 51, 63, 65, 99, 100]. Since the idea is to create a strong magnetic field from an external source and miniaturize the IPM embedded in the CE, having one cylindrical magnet as EPM would not be sufficient to compensate for the shrinkage of the IPM. Therefore, a strategy based on multiple magnets (including arrays of magnets) can enhance the magnetic field [88, 89]. These arrays can be analyzed by using the superposition principle, thus summing the contributions from all of the individual EPMs [86, 87]. Different shapes can be adopted in the array such as ring-shaped permanent magnets (as presented in this thesis) or cubic permanent magnets. For instance, we can begin by using two and four cylindrical permanent magnets diametrically magnetized as the source of the external magnetic field as shown in Fig. 3.16.
Figure 3.16 The centre of the system, called point P, coincides with the IPM’s centre, and is located at the centre of a circle with a radius of 60 mm. (a) Two cylindrical magnets, EPM1 and EPM2, generate a rotating magnetic field at point P; (b) four EPMs create a stronger rotating magnetic field at point P.

Each EPM’s diameter is 50 mm, length of 70 mm, and a magnetization of 1 T along the diameter. The IPM’s diameter is 10 mm, length of 10 mm, a magnetization $m$ of 1 T along its diameter, and a volume of $7.85 \times 10^{-7}$ m$^3$. The centre of each EPM is fixed and they can only rotate around their own axial axes. Although the EPMs could rotate towards any direction, we had to constrain their rotation about their axial axes (i.e., about the Z axis) for simplicity. However, the methodology and analysis of the magnetic interactions can be applied if different rotational directions are selected. It is also assumed during the simulations that the IPM’s centre coincides with the centre of the system (point P) and that the IPM can neither rotate nor translate, thus its magnetization vector $m$ remains constant in magnitude and orientation. Under these conditions, the coordinate systems XYZ and $X'Y'Z'$, the latter which is shown in Fig. 3.2, coincide. Therefore, the magnetic torque $\tau_z$, imparted to the IPM, as defined in Section 3.2, can be simply expressed as $\tau_z$. Fig. 3.17 shows an example of two EPMs rotating in the counterclockwise direction while IPM is fixed at point P.
Figure 3.17 Top view of the rotation in the counterclockwise direction of two EPMs is represented through the sequences (a), (b), (c) and (d). The IPM is fixed through all the sequences.

At point P, the magnetic flux density $B$ produced by two EPMs is the sum of the individual contributions of each EPM. If both EPMs are correctly synchronized as shown in Fig. 3.17, then the norm of $B$ will be two times the norm of the magnetic flux density produced by a single EPM.

A vector representation could be used to illustrate the change in the direction of the $B$ at point P as shown in Fig. 3.18. A direct comparison between Figs. 3.17-3.18 shows that while the EPMs rotate in one direction, the $B$ rotates in the opposite direction. According to Eq. 3.1, the maximum magnitude of the torque $\tau_x$ exerted on the IPM is produced in Fig. 3.18 (b) and in Fig. 3.18 (d) when the angle between $B$ and $m$ is $+90$ degrees.

Figure 3.18 The $B$ produced by EPMs rotates in the clockwise direction at point P where the IPM is located. The magnetization of the IPM $m$ is maintained constant since the IPM is fixed in the simulations.
3.5.3 Magnetic interactions

We are interested in calculating the magnetic torque $\tau_z$ exerted on the IPM, thus the only components of interest of the magnetic flux density are $B_x$ and $B_y$. For this purpose, we use the FEM solution COMSOL. If the radial distance is maintained under similar dimensions of the EPM, then the norm of $\mathbf{B}$ ($B_{\text{norm}}$) produced by a single cylindrical EPM at any radial distance can be approximated, by curve fitting, as follows

$$B_{\text{norm}} = |\mathbf{B}| = \sqrt{B_x^2 + B_y^2} = B_{\text{avg}} + B_{\text{peak}} \cos(\emptyset) \quad (3.14)$$

$B_{\text{norm}}$ produced by one EPM at a radial distance of 60 mm is shown in Fig. 3.19 when the EPM rotates an angle $\theta_{\text{EPM}}$ around its own axis. For this particular case, $B_{\text{norm}}$ estimated in the FEM software is given by

$$B_{\text{norm}} = 0.0648 + 0.0096 \cos(2 \theta_{\text{EPM}}) \quad (3.15)$$

where

$$\emptyset = 2 \theta_{\text{EPM}}$$

and $\theta_{\text{EPM}}$ is the angle of rotation of the EPMs.

Figure 3.19 Comparison of the magnetic flux density norm $B_{\text{norm}}$ simulated at a radial distance of 60 mm for single and multiple EPMs.

Figure 3.19 shows that when the magnetic system is made up of two and four EPMs; the magnitude of $B_{\text{peak}}$ decreases and the magnitude of $B_{\text{avg}}$ increases two times.
approximately. When the radial distance is increased up to 240 mm, Eq. 3.14 is no longer valid, as shown in Fig. 3.20. However, $B_{avg}$ decreases inversely with the third power of the distance as estimated in Eq. 3.4.

The relationship between the magnetic torque exerted on the IPM and the magnetic flux density created by the EPMs is given by Eq. 3.1. Since the IPM is fixed at the centre of the system, the only variable that is changing while the EPMs rotate is $B$.

The z-component of the magnetic torque imparted to the IPM could be expressed as

$$
\tau_z = \frac{V}{\mu_0} |\mathbf{m}| |\mathbf{B}| \sin(\gamma) 
$$

(3.16)

where $\gamma$ represents the angle between $\mathbf{B}$ and the IPM’s magnetization vector $\mathbf{m}$. Since the volume of the IPM is $7.85 \times 10^{-7}$ m$^3$, $\mu_0$ is $12.56 \times 10^{-7}$ N/m$^2$, $|\mathbf{m}|$ is 1 T, then this equation can be written as

$$
\tau_z = 0.625 B_{norm} \sin(\gamma) 
$$

(3.17)

Fig. 3.21 shows the variation of the angle $\gamma$ when four EPMs, which are magnetized as shown in Fig. 3.16 (b), rotate around the axis Z. This curve is the same when $\mathbf{B}$ is generated by one or two EPMs regardless of the radial distance. According to Fig.
$3.21, \gamma = -\theta_{EPM}$ for $0^0 \leq \theta_{EPM} < 180^0$ and $\gamma = \theta_{EPM}$ for $180^0 \leq \theta_{EPM} \leq 360^0$. This indicates that $\mathbf{B}$ rotates in the clockwise direction while the EPMs rotate in the counterclockwise direction as shown in Fig. 3.18.

![Angle of $\mathbf{B}$ with four rotating EPMs](image)

Figure 3.21 Angle of $\mathbf{B}$, $\gamma$ [deg], at a radial distance of 240 mm when four EPMs rotate around their own axes.

Using Eq. 3.16, we have estimated $\tau_z$ on the IPM at different radial distances as shown in Fig. 3.22.

![Magnetic torque $\tau_z$ exerted on IPM by four rotating EPMs](image)

Figure 3.22 Comparison of the magnetic torque $\tau_z$ produced on the IPM by four EPMs that rotate 360 degrees at different radial distances.
It is also possible to keep increasing the maximum torque at a radial distance of 240 mm if more EPMs are added to the system at adequate angles and with appropriate magnetization directions. For instance, six and eight cylindrical EPMs can be included in the entire system external magnetic system to obtain a higher $B$ as shown in Fig. 3.23. By adding two EPMs to the first four EPMs at 45 and 225 degrees respectively; and magnetized in the $+Y$ direction, it is possible to obtain a maximum torque of 12.1 mNm. In a similar way, two additional EPMs magnetized in the $-Y$ direction can be incorporated at 135 and -45 degrees, respectively. In this case, the maximum torque obtained will be 15.8 mNm. Fig. 3.24 shows the variation of $\tau_x$ at a radial distance of 240 mm for multiple EPMs. The maximum torque produced by four EPMs at this distance is 6 mNm. It can be seen that when the number of EPMs is doubled from four to eight, the magnetic torque increases slightly more than twice.

![Figure 3.23 Adding EPMs to increase $B$. (a) Two EPMs added at 45 and 225 deg whose magnetization are along the Y axis, (b) Two other EPMs magnetized in the $-Y$ direction.](image-url)
Figure 3.24 Magnetic torque $\tau_z$ produced by multiple EPMs at a radial distance of 240 mm.

Table 3.1 summarizes the results obtained for the magnetic torque exerted on the IPM by multiple EPMs at different radial distances.

<table>
<thead>
<tr>
<th>Radial Distance [mm]</th>
<th>One EPM</th>
<th>Two EPMs</th>
<th>Four EPMs</th>
</tr>
</thead>
<tbody>
<tr>
<td>60</td>
<td>45.0</td>
<td>70.0</td>
<td>160.0</td>
</tr>
<tr>
<td>120</td>
<td>8.0</td>
<td>16.0</td>
<td>30.0</td>
</tr>
<tr>
<td>180</td>
<td>4.1</td>
<td>7.2</td>
<td>10.8</td>
</tr>
<tr>
<td>240</td>
<td>2.8</td>
<td>4.9</td>
<td>6.0</td>
</tr>
</tbody>
</table>

In order to miniaturize the IPM, we have decided to simulate an IPM with half of the volume of the original one by reducing its length from 10 mm to 5 mm. We have placed the IPM at a radial distance of 60 mm for one, two and four EPMs and we have obtained maximum torques of 25.0, 35.0 and 80.0 mNm, respectively. These results are half of the torques listed in Table 3.1 for an IPM’s length of 10 mm. These results are in agreement with Eq. 3.16 and show a linear correlation between the IPM’s volume and the magnetic torque.
3.5.4 The slider-crank mechanism

The IPM is attached to the crank of a slider-crank mechanism as shown in Fig. 3.15. When the IPM is rotated by the EPMs, the piston will generate a rectilinear movement and the force generated by the piston will be used to release the drug from the reservoir.

Equations 3.8-3.13 describe the kinematics and kinetics of the slider-crank mechanism and can be used to design and optimize the DDS for CE. For instance, making the length of the crank \( R \) equal the radius of the cylindrical IPM, 5 mm, and choosing the length of the connecting rod \( L \) as 15 mm, we obtain the maximum distance tolerated by the CE as \( 2*R+L=25 \) mm when \( \alpha \) is \( 0^\circ \) (using Eq. 3.8). This distance equals the length of a commercial CE. When \( \alpha \) is \( 180^\circ \), we obtain \( x=10 \) mm, allowing a maximum distance of 15 mm in the drug reservoir. This space will allow accommodating a maximum drug volume of 1.18 mL. This amount of drug is slightly higher than the maximum amount of drug volume of 1 mL reported in [66].

In regard to the requirements for the crankshaft torque, we can estimate it by using the model presented in Section 3.4 as follows. \( \varphi_c \) is maximum when \( \alpha \) is \( 90^\circ \). At this crank angle, \( \varphi_c \) is \( 19.48^\circ \). Since the analysis is usually made at a constant angular velocity [96], then the angular acceleration of the crank \( \dot{\alpha} \) is always 0. Consequently, the moment of inertia of the crank, \( I \), will not contribute to Eq. 3.13. The angular velocity \( \omega \) could vary from 20 to 120 rpm (i.e. from 2 to 13 rad/s) [9]. When \( \omega \) is maximum, we find the piston position and acceleration are 14.1 mm 0.2817 m/s\(^2\), respectively. In [12], the total weight of the capsule prototype is 4.2 g and it includes four small cylindrical magnets. Therefore, we can assume that the maximum weights combine \( m_{lb} \) and \( m_p \) which is 3 g. Using these values, we find the maximum force \( F_d \), described by Eq. 3.12, is 0.845 mN. The piston force \( P \), which is shown in Fig. 3.15, could be assumed as representing 90% of the required anchoring force, which is 912 mN [6, 8]. If \( P \) is higher than the anchoring force, the capsule may detach from the intestine wall. Replacing all these values in Eq. 3.13, we estimate the maximum crankshaft torque \( \tau_c \) as 4.1 mNm. Therefore, the magnetic torque transmitted from the EPMs to the IPM should be equal or less than 4.1 mNm to effectively achieve a maximum piston force \( P \) of 820 mN to release the drug.
The contribution of the force $F_d$ is maximum when $\omega$ is 13 rad/s. However, even at this maximum angular speed, its contribution of 0.845mN to the crankshaft torque can be neglected. Thus, the maximum torque to be exerted on the crank should be about 4 mNm. By neglecting $F_d$ and the moment of inertia of the crank $I$, we can simplify the crank shaft torque given in Eq. 3.13 as

$$\tau_c = P_x \tan \varphi_c \quad [\text{Nm}] \quad (3.18)$$

which can be rewritten as

$$\tau_c = PR \sin \alpha (1 + \frac{R}{L} \cos \alpha) \quad [\text{Nm}] \quad (3.19)$$

This is the crankshaft torque $\tau_c$ to balance the discharge force $P$ [96]. This torque can be provided by two or four EPMs at radial distances between 180 and 240 mm from the IPM centre (see Table 3.1). We estimate that a force $P$ of 820 mN, which equals a pressure of 10.45 kPa on a circular orifice surface of 78.5 mm$^2$, is enough to release a wide variety of drug compounds.

### 3.5.5 Results and discussion

We show that the external magnetic flux density can be enhanced at a specific region in the space when multiple cylindrical EPMs are added at the appropriate angle and with the correct magnetization direction. This design optimization of the EPMs improves the magnetic flux density and help to compensate the loss of magnetic link when the IPM shrinks. In our particular case of WCE, the miniaturization of the IPM provides more volume for the drug reservoir and other components. When the IPM’s volume is reduced by 50%, it is required to double the number of cylindrical EPMs to compensate the loss and obtain a similar magnetic torque. Furthermore, the addition of EPMs in the system, at optimal angular positions and with adequate magnetization directions, is also useful to compensate the loss of magnetic link caused by increments in the distance between EPMs and the IPM.

The results reported in this section suggest that two and four cylindrical EPMs at radial distances between 180 and 240 mm are enough to exert a magnetic torque higher than 4 mNm on a cylindrical IPM (diameter: 10 mm and length: 10 mm). This torque is transmitted to the piston that exerts a piston force of 820 mN. An improvement in
the magnetic linkage can also be obtained by optimizing the IPM embedded in the capsule robot and this is presented in the next section (Section 3.6).

3.6 Optimization of the IPM

The tight volume constraints imposed by the capsule size and the lack of power to actuate all those components within the robotic capsule are challenges that require careful consideration in its design, optimization and development. Specifically, magnetic actuation has been successfully used in several prototypes of robotic endoscopic capsules to actuate wirelessly a variety of on-board mechanisms to overcome the limitation of scarce energy available within the robotic capsule [9, 10, 12, 63, 65, 75, 76]. However, further miniaturization and optimization of the on-board permanent magnets (i.e., the IPMs) are required to obtain an efficient magnetic linkage (i.e., an optimized magnetic force or torque imparted to the IPM).

In Section 3.5 we have focused on the design optimization of the external magnetic system [101]. However, according to Eq. 3.1, the magnetic torque transmitted to the IPM depends not only on the external magnetic field but also on the type of IPM placed within the capsule robot. Therefore, in this section, we focus on the size and shape optimization of the IPM. To this end, we use analytical solutions which are more efficient for the design and optimization of magnetic systems than the time consuming finite-element methods [84].

Specifically, we aim to compare the magnetic torque transmitted to cubic and cylindrical IPMs which are subject to the same external magnetic field. Therefore, in this section (Section 3.6), we focus on the optimization of the IPM and relatively less attention is given to the external magnetic system in this section. We choose these two different shapes because they are the most commonly used for magnetic actuation in medical applications [9, 10, 12, 63, 65, 75, 76, 100]. However, the analytical solutions used in this section are not limited to only these two shapes of IPMs and can be used with any other shape or size without loss of generality.
3.6.1 Magnetomechanical system

A small cubic or cylindrical IPM, as shown in Fig. 3.25 (a), is to be placed in a prototype of capsule robot. The IPM is driven by the rotating magnetic flux density $\mathbf{B}$ that is created as the array of twelve ASMs is rotated about the Z axis as shown in Fig. 3.25 (b). This magnetic system made of 12 ASMs has been optimized to generate approximately 303 mT at the centre of the system as it is presented in Section 5.1. The rotating $\mathbf{B}$ can be also created by an array of cylindrical EPMs (as presented in Section 3.5). However, the external magnetic system, its optimal geometric configuration and optimal shapes are not the focus of the analysis in this section but are presented in Chapters 4 and 5.

![Figure 3.25](image)

(a) (b)

Figure 3.25 (a) A cylindrical IPM whose centre coincides with the centre of the system (called point P) and is actuated by an array of 12 ASMs, (b) the ASMs generate approximately $|\mathbf{B}| = 303$ mT at the centre of the system.

The centre of the system, called point P, is located at the centre of a circle with a radius of 30 mm. The internal and external radii of each ASM are 30 mm and 50 mm, respectively. The height of each segment is 30 mm and the magnetization grade is 1.32 T (i.e., N45). The dimensions of this prototype of external magnetic system were selected only to prove the feasibility and operation of the actuation system. However, the magnetic system can be scaled up to surround the patient's body as presented in Chapter 4. As the optimization of the IPM that is carried out in this section is valid for any size or shape of permanent magnets, we choose cubic and cylindrical IPMs for practical reasons.
3.6.2 Theoretical methods

The magnetic torque $\tau$ imparted to the IPM by the 12 ASMs is described by Eq. 3.1. This analytical model, which we call Model1, is commonly used to estimate the magnetic torque imparted to IPMs in prototypes of WCE to actuate a variety of mechanisms [9, 10, 76, 79]. In this analytical model, $\mathbf{B}$ represents the magnetic flux density computed at the centre of the IPM that has a volume of $V$. Although calculating $\mathbf{B}$ at the centre of the IPM can ease the analysis of the magnetic torque and decrease the computation time, two difficulties are present when using this Model1. Firstly, this analytical model does not indicate if there is any difference in the torque imparted to a cubic or a cylindrical IPM under similar conditions (i.e., assuming that the volume $V$ and magnetization grade of the IPM are the same and that $\mathbf{B}$ is not changed). Secondly, Model1 only allows the computation of the magnetic torque about the main axes of the IPM (i.e, the axes $X', Y'$, and $Z'$ shown in Fig. 3.2) but cannot be used to compute the torque with respect to any other axes.

In order to overcome these drawbacks present in Model1, we use another analytical model to compute $\tau$, which we call Model2 and is given by [86]

$$\tau = \mathbf{r} \times \mathbf{F} \quad [\text{Nm}]$$

(3.20)

where $\mathbf{F}$, which is described by Eq. 3.7, is the magnetic force exerted on the IPM with a volume $V$ that is exposed to $\mathbf{B}$, and $\mathbf{r}$ is the vector from the fulcrum to the point where $\mathbf{F}$ is applied. In Model2, $\mathbf{B}$ is not calculated only at the centre of the IPM but also on the IPM’s surfaces and through its volume as it is expressed by Eq. 3.4. For this reason, the computation time of Model2 is greater than the computation time of Model1.

Since Model2 allows the computation of the magnetic torque around any axis, and therefore represents a more general model compared to Model1, we use it to compute $\tau_{z'}$ (i.e., the transmitted torque about the $Z'$ axis) on the cubic and cylindrical IPMs as follows. Firstly, we make the IPM’s centre to coincide with the centre of the system and we also align its magnetization vector $\mathbf{m}$ with the $X$ axis for simplicity as shown in Fig. 3.26. Therefore, the coordinate systems $XYZ$ and $X'Y'Z'$, the latter which is shown in Fig. 3.2, coincide. In this case, the magnetic torque $\tau_{z'}$ imparted to the IPM, as
defined in Section 3.2, can be simply expressed as $\tau_z$. Secondly, $m$ remains aligned with the $X$ axis at all times as we rotate the external magnetic system by increments of $30^0$ and compute $\tau_z$, using both Model1 and Model2, until the external magnetic system completes a full rotation of $360^0$.

Figure 3.26 Cylindrical IPM diametrically magnetized. Its magnetization vector $m$ is fixed and aligned with the $x$ axis as the external magnetic system completes a $360^0$ rotation.

A comparison of the torque transmitted to these two IPMs is possible by assuming that the volume and magnetization grade of each IPM are the same. Therefore, we choose for practical reasons a 3.175 mm cubic IPM with a magnetization grade of 1.4 T (i.e., N50). For the cylindrical IPM diametrically magnetized, we choose its length $L$ to be 3.175 mm and find its radius $R$ to be 1.79 mm (i.e., $L/\sqrt{pi}$) and its magnetization grade is also 1.4 T. With these specifications for both IPMs which are subject to the same rotating magnetic field, we guarantee that an appropriate comparison can be carried out in regard to the $z$ component of the magnetic torque exerted on them individually. Fig. 3.27 shows the results of this comparison where $\theta_{EPM}$ represents the angle by which the external magnetic system is rotated. $\theta_{EPM}$ is the same misalignment angle between $m$ and $B$ because $m$ remains aligned with the $X$ axis as the external magnetic system rotates.
Figure 3.27 (a) Z component of the magnetic torque imparted on cubic and cylindrical IPM using Model1, (b) Z component of the magnetic torque imparted on cubic and cylindrical IPM using Model2.

We can see from the theoretical results of Model1 that the same peak torque of 11.04 mNm is imparted on either a cylindrical or a cubic IPM-they perfectly match each other. However, the theoretical results of Model2 show that a peak torque of 13.82 mNm is exerted on the cylindrical IPM while a peak torque of 10.87 mNm is transmitted to the cubic IPM. This difference may be even bigger if the volume of the IPM is increased or if the magnitude of the magnetic field is increased. For example, Fig. 3.28 shows the comparison of the peak torque transmitted to cylindrical and cubic IPMs as their volume is increased. Since the diameter of a WCE is typically 11 mm [66], we choose a maximum length L of 10 mm for the results in Fig. 3.28 to make sure that the IPM can fit in the capsule. In order to guarantee that the cylindrical IPM’s volume is equal to the volume of the cubic IPM, we also choose its radius R to be $L/\sqrt{\pi}$.

Figure 3.28 Comparison of the peak torque when the volume of the IPM is increased.
These results from Model2, which cannot be predicted by Model1, indicate that it is more efficient to transmit a magnetic torque on a cylindrical IPM diametrically magnetized than on a cubic IPM if they have the same volume and if they are placed at the centre of the system. Furthermore, by choosing L between 4 mm and 5 mm, it is possible to obtain differences in the torque transmitted up to $\Delta \tau_z = 11 \text{ mNm}$ between a cylindrical IPM and a cubic IPM. For example, if L=5 mm, the peak torque imparted on the cylindrical IPM is 53.67 mNm while on the cubic IPM is 42.44 mNm. This additional torque of about $\Delta \tau_z = 11 \text{ mNm}$ provided by the cylindrical IPM represents a significant amount that would allow the actuation of other on-board mechanisms if we consider that a magnitude of 5.3 mNm is used in [12] to actuate a biopsy mechanism in WCE and a magnitude of about 5 mNm is used in [76] to deploy legs in WCE for locomotion purposes.

Although this comparison is carried out by placing the IPMs at the centre of the system, a similar comparison for the transmitted torque can be conducted if the IPMs are located at other positions and orientations since the analytical models are general. However, in this section (Section 3.6), we are only interested in actuating an IPM that is to be placed at the centre of the system. Chapters 6 and 7 are dedicated to the analysis of the transmitted torque to an IPM with arbitrary position and orientation. Therefore, the experimental results in the next subsection are for a cylindrical IPM whose centre coincides with the centre of the system. It must be noted that the magnetic torques estimated for the cylindrical IPM using Model 1 and Model 2 are different, as shown in Fig. 3.27. The aim is to demonstrate that Model 2 accurately estimates the magnetic torque by taking into account the geometry of the magnet, rather than purely considering what its volume is.

**3.6.3 Experimental methods**

**3.6.3.1 Magnetic torque**

In order to validate the accuracy of Model2, we have decided to use a cylindrical IPM axially magnetized whose centre was shifted along the X axis as shown in Fig. 3.29. We use a cylindrical IPM with a diameter of 3.12 mm (i.e., $R=3.12/2 \text{ mm}$), length L of 6.24 mm and magnetization grade of 1.32 [T] (i.e., N45).
With reference to Fig. 3.29, we measured the z components of the magnetic torques about the centre of the system, called $\tau_z$, and about the centre of the IPM, called $\tau_{zr}$. We fabricated two plastic connectors, using a 3D printer, to hold the IPM as shown in Fig. 3.30. Thus, the plastic connector1 and plastic connector2 were used to measure $\tau_z$ and $\tau_{zr}$, respectively. A torque gauge (HTG2-40 supplied by IMADA) with its respective torque sensor held the IPM at $x_1=15.6$ mm (and $y=z=0$). The torque sensor can be moved along the X and Z axes and the arrays of the magnets can only be moved along the Y axis. These displacements are controlled by a micromanipulation system based on an X-Y-Z stage.

![Figure 3.29 A cylindrical IPM axially magnetized whose centre is shifted along the X axis to $x=x_1$. Its magnetization vector $m$ is fixed and aligned with the X axis as the external magnetic system completes a $360^\circ$ rotation.](image)

![Figure 3.30 Plastic connectors to measure $\tau_z$ and $\tau_{zr}$.](image)

The experimental results for the magnetic torque imparted to this cylindrical IPM about the centre of the system $\tau_z$, and about the centre of the IPM $\tau_{zr}$, are shown in Figs.
3.31-3.32, respectively. Please note that it is not possible to estimate $\tau_z$ by using Model1, since Model1 only allows the computation of the magnetic torque about the centre of the IPM, thus only the theoretical results of Model2 are shown in Fig. 3.31.

![Magnetic Torque $\tau_z$](image)

**Figure 3.31** $Z$ component of the magnetic torque about the centre of the system imparted on a cylindrical IPM axially magnetized whose centre is shifted to $x_1=15.6$mm.

![Magnetic Torque $\tau_z'$](image)

**Figure 3.32** $Z'$ component of the magnetic torque about the centre of the IPM imparted to a cylindrical IPM axially magnetized whose centre is shifted to $x_1=15.6$mm.

The experimental results for the magnetic torque imparted on this cylindrical IPM when its centre coincides with the centre of the system (i.e., $x_1=0$) are shown in Fig. 3.33.
Figure 3.33 Z component of the magnetic torque about the centre of the IPM imparted to a cylindrical IPM axially magnetized whose centre coincides with the centre of the system.

The theoretical results of Model2 for $\tau_z$, which cannot be predicted by Model1, are in agreement with the experimental results as shown in Fig. 3.31 for the torque transmitted to the IPM about the centre of the system. Furthermore, if the IPM’s centre coincides with the centre of the system, then both models estimate the same results when computing the torque about the centre of the IPM as shown in Figs. 3.32-3.33. In other words, when $x_1=0$, $\tau_z = \tau_{zr}$. Therefore, these experiments validate the accuracy of Model2. Since we are only interested in actuating an IPM that is to be placed at the centre of the system, either model can be used to estimate the torque transmitted about the Z axis. We used this cylindrical IPM axially magnetized and placed it at the centre of the system to actuate a slider-crank mechanism and the details of this mechanism and the experimental results for it are presented in the next subsection.

### 3.6.3.2 The slider-crank mechanism

We fabricated the slider-crank mechanism from a plastic material (ABS) with a 3D printer. All its components are depicted in Fig. 3.34. The IPM’s centre coincides with the centre of the system, thus $\text{XYZ}$ and $\text{X’Y’Z’}$ are aligned. Since the IPM is connected to the crank of the slider-crank mechanism, the piston will release the drug from the reservoir when the IPM is rotated around the Z axis by the external magnetic field (see Fig. 3.15).
Figure 3.34 (a) The cubic IPM case connected to a disk through the crankshaft; (b) components of the slider-crank mechanism. A. Platform, B. Connecting rod, C. Piston, D. Laser reflective surface, E. Spring holder, F. IPM case, G. Platform supporter. β is the angle formed between the external magnetic system and the X axis \(180^\circ<\beta<-180^\circ, \beta=180^\circ-\theta_{\text{EPM}}\) and \(\theta_{\text{EPM}}\) is shown in Fig. 3.1.

Due to practical reasons, we placed a helical spring in the slider-crank mechanism to measure the piston force \(F_s\) by using the Hooke’s law:

\[
F_s = K \Delta x = K(x - x_{\text{min}}) \tag{3.21}
\]

\(\Delta x\) represents the displacement of the spring and \(K\) is the stiffness of the spring which was measured as 1.59 N/mm. In our experiments, we manually rotated the arc-shaped magnets and the cylindrical IPM rotated the crank at the same time, compressing the spring when \(\alpha\) (i.e., the crank angle defined in Section 3.4) changed from \(180^\circ\) to \(0^\circ\) and extending the spring when \(\alpha\) changed from \(0^\circ\) to \(-180^\circ\). A laser (optoNCDT 1700 by Micro-Epsilon) was used to measure the stroke \(x\) as shown in Fig. 3.35. The beam of the laser reflects on the reflective surface that is connected to the piston as shown in Fig. 3.34 (b). The laser reading was used to estimate \(\alpha\) and \(F_s\) given by Eq. 3.10 and Eq. 3.21, respectively. Once \(F_s\) and \(\alpha\) are estimated, we use Eq. 3.19 to estimate the torque delivered to the crankshaft by the force \(F_s\) (Note: \(P=F_s\) when using Eq. 3.19). With reference to Fig. 3.15, the slider-crank mechanism was fabricated with the dimensions of \(R=3\) mm and \(L=9\) mm.
Figure 3.35 Experimental setup to measure the piston force $P = F_s$ and the crankshaft torque $\tau_c$ in the slider-crank mechanism.

Figure 3.36 The experimental piston force $P = F_s$ under the external magnetic field.

Figure 3.36 shows the piston force generated as the external magnetic system rotates one full cycle, compressing the helical spring in the left hand side of the curve and extending it on the other half of the curve. The experimental peak force of 4.6 N was obtained when the external magnetic system reached $\beta = -30^0$. At this point, we estimate $\alpha = 82^0$ and the misalignment angle between $m$ (the IPM magnetization direction) and the direction of the external magnetic field (i.e., $\theta_{EPM}$ with reference to Fig. 3.33) was $112^0$, producing the peak torque of 14 mNm, shown in Fig. 3.37. This peak torque of $\tau_c$ estimated with the laser reading is the same magnetic torque exerted on the IPM (i.e., $\tau_x$) and measured with the torque sensor when $\theta_{EPM} = 112^0$ as shown in Fig. 3.33, thus it validates our results and Model2, which is accurate enough to optimize the size of the IPM.
3.6.4 Results and discussion

The optimization of the on-board components embedded in a robotic capsule is critical due to the tight volume constraints imposed by the capsule size and the lack of power for the actuation of such components. In this section, we tackle these two critical issues by focusing on the size and shape optimization of the IPM to be embedded in a capsule robot for drug delivery.

We compare the torque transmitted to the small cubic and cylindrical IPMs that could be embedded in the existing wireless capsule endoscopes. To this end, we use two different analytical models, Model1 and Model2, and carried out experiments to verify their accuracy. We find that, under the same external magnetic field and assuming that each IPM has the same volume and magnetization grade, a cylindrical IPM diametrically magnetized always provides a higher magnetic torque than a cubic IPM. Furthermore, this efficacy in torque transmission becomes more evident (or it rapidly increases) as the volume of the IPM is increased. These results suggest that an optimal volume for a cylindrical IPM can be selected to actuate multiple on-board mechanisms which are to be included in the next generation of WCE.

In our experimental section (Section 3.6.3.2), we connected a cylindrical IPM to a slider-crank mechanism to measure the torque exerted on the IPM as the external magnetic system was manually rotated. This magnetic torque was converted into a
piston force that would expel drug out of the reservoir. A peak torque of about 14 mNm was converted into a peak force of about 4.6 N which is more than enough to release a variety of drug compounds if we consider that a peak piston force of only 820 mN is required for drug delivery [101].

3.7 Conclusions

We have carried out the design optimization of the external magnetic system and the size and shape optimization of the IPM. These results indicate that the optimization of the magnetic linkage is useful to increase the operating distance and/or miniaturize the IPM that is to be embedded in the capsule robot.

Although, in this thesis, the slider-crank mechanism is not optimized to improve the torque or its volume is not optimized within the capsule robot, we suggest that other mechanical designs such as a cam mechanism or a yoke mechanism can be studied, in the future, as potential solutions to fabricate more compact mechanical systems. Therefore, we focus more on the magnetic system in the rest of this thesis. We continue with the shape optimization of the external magnetic system to enhance the magnetic field in Chapter 4.
Chapter 4
Shape Optimization of the External Magnetic System

In Chapter 3, we have shown that the miniaturization of an active drug delivery mechanism in the WCE is possible when the external magnetic field is properly applied and also if an appropriate shape for the IPM is chosen [101, 102]. In this chapter, we report on the design and shape optimization of the external magnetic system which can be realized by an array of permanent magnets. This optimization allows large operating distances and does not impose strict design constraints on the miniaturization of the components inside a WCE.

4.1 Objectives and limitations

Our proposed DDS for WCE uses different mechanisms and modules (see Fig. 3.1) to achieve on-demand drug release. Although each module and component could be optimized to improve the overall system, in this chapter, we mainly focus on the optimization of the magnetic interactions between the external magnetic system that is made of external permanent magnets (EPMs) and the IPM.

In a real application, the robotic capsule is free to move and rotate within the cylindrical volume of radius d and length L of the external magnetic system which is shown in Fig. 3.3. However, in order to facilitate the analysis of the magnetic interactions between the EPMs of the external magnetic system and the IPM, we introduce the following specific physical constraints on these permanent magnets. First, we align the two coordinate systems XYZ and X_aY_aZ_a (see Fig. 3.1 and Fig. 3.3 for the definition of both coordinate systems) by coinciding the center of the external magnetic system with point P and by putting in parallel planes XY and X_aY_a (i.e., the external magnetic system is not inclined, thus \( \varphi=0^0 \)). Therefore, the ring-shaped external magnetic system can only rotate about the Z axis (i.e., \( 0^0 \leq \theta_{EPM} < 360^0 \)). Nevertheless, in the real application, the external magnetic system's location and orientation could be controlled by adjusting its center and its angles \( \theta_{EPM} \) and \( \varphi \) (defined in Fig. 3.1) with the joystick and a transformation matrix can be easily used when working with XYZ
and X_aY_aZ_a reference systems. Second, although the IPM’s axial axis (Z') can also be inclined with respect to the Z_a axis of the external magnetic system, we constrain it in this chapter to be always parallel to Z_a. The analysis of the transmitted torque \( \tau_{z'} \) when the IPM’s axial axis is tilted is presented in Section 6.2 and Chapter 7. Third, we also make the plane X'Y' (defined in Section 3.2) of the IPM coincide with the plane X_aY_a of the external magnetic system, so that axial movements of the IPM are not part of this chapter but are later discussed in Section 6.1.

Finally, since the ring-shaped external magnetic system is symmetrical in the Z axis, we choose half of the length L to be the plane z=0 as the plane on which the IPM’s center moves within a circle of radius d centered at point P. With these constraints, we only allow the movement of the IPM’s center on the plane z=0 with the IPM’s axial axis parallel to the Z axis, and subsequently \( \tau_{z'} \) can be simply denoted as \( \tau_z \) which is expressed in Eq. 3.16. Keeping in mind that the magnetic flux density B created by the EPMs decreases with the distance and point P is located at the furthest distance from the inner surface of the external magnetic system (on plane z=0), then point P represents a critical point for B and also for \( \tau_z \). Therefore, we aim to increase B at point P and also the magnetic torque imparted to the IPM, the center of which is located at that critical point P. Higher B and magnetic torques are then expected at any other point within the circle with a radius of d where the IPM’s center can be placed. If the IPM’s center moved axially or radially away from point P, then the position and orientation of the external magnetic system can be controlled from the joystick to obtain an adequate alignment between the EPMs and the IPM.

Finding an optimal configuration and shape of the EPMs within the ring of the external magnetic system is crucial to improve the magnetic torque that is transmitted to the IPM, and this can allow the miniaturization of the IPM’s size and increase the operating distance of the DDS at the same time. Therefore, we present the design and shape optimizations of the EPMs to increase B at point P. This process is carried out using analytical models.
4.2 Magnetic field analysis

4.2.1 Magnetic systems consisting of arrays of multiple permanent magnets

The magnetic torque $\tau_z$, which is described by Eq. 3.16, is useful to analyze the effect of changes in the magnitude of the flux density on the transmitted torque. Since the magnetic torque is proportional to $|B|$, we aim to enhance $|B|$ at the critical point P. It should be noted that only $B_x$ and $B_y$ components of $B$ contribute to $\tau_z$. Although the magnetic torque is also proportional to the IPM’s volume $V$, variations in the IPM’s dimensions and their effects on the magnetic torque and piston force are presented in the experimental section (i.e., subsection 4.4.3).

In our previous work [101], we have used a FEM solution (i.e., Comsol) and showed that multiple EPMs can be used to create a stronger $|B|$ than the one produced by a single EPM at point P. We also presented, in our previous work, that multiple EPMs (up to 8) arranged along a circle at appropriate locations and with certain orientations can impart higher magnetic torques to a small IPM. In this section, we perform parametric studies to determine a suitable array of permanent magnets to be placed in the ring-shaped external magnetic system shown in Fig. 3.1 by using analytical models since these studies are extremely time-consuming with FEM methods [84].

The first type of array consists of diametrically magnetized cylindrical magnets and the second type consists of arc-shaped permanent magnets (ASMs), as shown in Fig. 4.1. Although cylindrical magnets are commonly used in medical applications [9, 12, 63, 65, 75, 76], ASMs have shown promising results in the transmission of high density torques as reported in [103, 104]. Therefore, we consider them in this study.

In order to understand the magnetic flux density produced by these arrays of permanent magnets, we analyze the contribution of each magnet, $i$, and then use the superposition principle to obtain the total magnetic flux density $|B|$.

By using the magnetic charge model [86], we calculate $B$ using Eqs. 3.4-3.6 which are general and can be used for any shape and size of magnets. We take $B$ as the total flux density generated by each type of array of permanent magnets. In particular, $|B| = \sqrt{B_x^2 + B_y^2 + B_z^2}$, but only the components of the magnetic field in the XY plane will
contribute to the magnetic torque about the Z axis, which is the axis of rotation of the IPM. We use the analytical models for B generated by a diametrically magnetized cylindrical magnet [105] and by ASMs [106], and also the superposition principle to find the total magnetic field generated by the arrays of magnets.

Figure 4.1 (a) A diametrically magnetized cylindrical magnet with radius R, length L1=2z2-z1 and magnetization grade M, (b) Arc-shaped permanent magnet, (c) top view of different types of arc-shaped permanent magnets (i.e., A1, A2, A3, and A4) used in this work.

4.2.2 Comparison of the magnetic flux density created by cylindrical and arc-shaped magnets

In the analysis of the magnetic flux density created by the external magnetic system, we align Xa and X axes (i.e., θ_{EPM} = 0°) because there is no need to rotate the external magnetic system. We start by placing one cylindrical EPM (radius R and length L, diametrically magnetized and magnetization grade M) at θ=180° with its center located at a radial distance of d+R from the center of the system (point P) as shown in Figs. 4.2 (a)-(b). We set a relatively large operating distance d of 240 mm and align M with the X axis to facilitate the analysis so that \( |B| = \sqrt{B_x^2 + B_y^2} \) at point P. We aim to create a B_x of 103 mT at point P since this value seems to be reasonable for the actuation of small IPMs [12].
We perform a parametric study using analytical models by varying L and R to find the minimum volume, $V_{\text{min}}$, of the cylindrical EPM that creates 103 mT from an operating distance $d$ of 240 mm. Figs. 4.2 (c)-(d) show that although there are many ways to create 103 mT from that distance (for example, a cylinder with $L=240$ mm and volume of approximately $5.7 \times 10^{-2}$ m$^3$ creates 103 mT, but also a cylinder with $L=600$ mm and volume of approximately $4.5 \times 10^{-2}$ m$^3$ creates 103 mT), there is a minimum cylindrical volume that generates 103 mT from the distance $d$. We find the following optimal parameters for a cylindrical EPM diametrically magnetized: $R_{\text{optimal}}=175$ mm, $L_{\text{optimal}}=425$ mm and $V_{\text{min}}=40.9 \times 10^{-3}$ m$^3$.

We replace this optimal cylindrical EPM with four cylindrical EPMs, each denoted as $C_i$ ($i=1,2,3,4$) and with 25% of the volume $V_{\text{min}}$ (i.e., $10.225 \times 10^{-3}$ m$^3$, $R=87.5$ mm, $L_1=425$ mm) and place $C_1$ at $\theta=180^0$, $C_2$ at $\theta=0^0$, $C_3$ at $\theta=90^0$, and $C_4$ at $\theta=270^0$ with magnetization directions as shown in Fig. 4.3 (a). We denote this configuration as $C_{1234}$. $B_x$ created by $C_{1234}$ at the center of the system is increased to 141 mT. Instead of using 4 cylindrical EPMs, if we use 4 ASMs each with a volume $V_{\text{ASM}}=V_{\text{min}}/4$, as shown in Fig. 4.3 (b), then $B_x$ at the center of the system will be 158 mT. The specifications of each ASM, which in this configuration we denote as $A_{1234}$, are: $r_1=240$ mm, $r_2=386.6$ mm, $L_2=425$ mm, angular width $\Delta \theta$ of $30^0$, 2 segments radially magnetized (i.e., $A_1$ and $A_2$) and 2 other segments tangentially magnetized (i.e., $A_3$ and $A_4$). Magnetization grade: 1.32 T. We also use analytical models and find that the positions of the ASMs presented in Fig. 4.3 (b) are optimal to increase $B_x$ at the center of the
system [107] and the details of this optimization process are presented in Section 5.1. The same optimal configuration with 4 cubic magnets is reported in [108], which indicates that for these shapes of EPMs, a maximum magnetic flux density is obtained at the centre of the system when the EPMs are arranged as shown in Fig. 4.3.

These results for $B_x$ generated at the centre of the system (also shown in more detail in Fig. 4.4) indicate that given the distance $d$ and the magnitude of $B_x$ that we want to generate, it is more efficient to distribute the volume of the EPMs along the circle with radius $d$ than to allocate the entire optimal volume $V_{\min}$ to a single cylindrical EPM. Fig. 4.4 compares $B_x$ along the X axis for one optimized cylindrical EPM (with optimized volume $40.9\times10^{-3} \text{ m}^3$), 4 cylindrical EPMs (each with a volume of $10.225\times10^{-3} \text{ m}^3$) and 4 ASMs (each with a volume of $10.225\times10^{-3} \text{ m}^3$). The optimization of location, orientation and shape of each EPM is of great importance when the magnetic system is scaled up since a minimum weight of the EPMs would be highly desirable not only to ease its maneuverability but also to reduce the costs associated with the fabrication of the EPMs.

Since, for the same minimum volume $V_{\min}$, an array of 4 ASMs produces the highest magnetic field at the center of the system where the IPM is located, we conduct
experiments with a scaled down structure that consists of 4 ASMs arranged in the configuration denoted as $A_{1234}$.

### 4.2.3 Scaled down external magnetic systems

For practical reasons, we choose an operating distance $d$ of 30 mm and conduct the inverse analysis: we initially estimate $B_x$ at point P generated by a given total volume $V_{\text{total}}$ that is equally divided into 4 ASMs (i.e., $A_{1234}$). This flux density value is then compared with $B_x$ created by the array $C_{1234}$ and also by a single cylindrical EPM. The combined volume of the whole array $C_{1234}$ equals $V_{\text{total}}$ and the volume of the single cylindrical EPM is also $V_{\text{total}}$ and the latter is placed at the position and orientation shown in Figs. 4.2 (a)-(b).

The operating distance $d$ of 30 mm, which represents an operating distance of 240 mm decreased by 8 times, is chosen due to the commercial availability of inexpensive ASMs with such dimensions that are used in our experimental section. The specifications of these permanent magnets are as follows: magnetization grade of 1.32 T (i.e., N45), $L_2=30$ mm, $\Delta \theta = \pi/6$, $r_2=50$ mm, $r_1=30$ mm, $V_A=12.564 \times 10^{-6}$ m$^3$ and $V_{\text{total}}=4 \times V_A$. The same total volume $V_{\text{total}}$ can be equally divided into 4 cylindrical EPMs (each cylinder with $R=11.55$ mm and $L_1=L_2$) the centers of which are located at a radial distance $d_1$ of 41.55 mm. These two arrays of magnets can be compared in terms of the $B_x$ generated at point P with a single cylindrical EPM (with a total volume $V_{\text{total}}$, $R=23.1$ mm and $L=L_2$) the center of which is located at the radial distance of $d+R=53.1$ mm. Fig. 4.5 shows the comparison of $B_x$ along the X axis generated by these scaled down external magnetic systems. The system $A_{1234}$ generates a $B_x$ of 113 mT at point P, while the structure $C_{1234}$ generates 103 mT and the single cylindrical EPM only generates 74 mT. The volume of the single cylindrical EPM would have to be increased to generate 103 mT at point P.

The comparisons of $B_x$ generated by these external magnetic systems, shown in Figs. 4.4-4.5, indicate that the structure $A_{1234}$ generates the highest $B_x$ at point P where the IPM is to be placed. If the IPM is moved along the X axis, it will be subjected to higher flux densities and consequently a higher magnetic torque can be transmitted to the IPM. For example, for some negative values of x, a single EPM produces a higher $B_x$. 

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but for positive values of $x$, the array $A_{1234}$ can produce higher flux densities. The arrays of magnets also produce a relatively constant value of $B_x$ over a larger region along the $X$ axis when compared with the $B_x$ produced by a single EPM, thus guaranteeing a more steady torque on the IPM.

![Flux Density Diagram](image)

**Figure 4.4** Comparison of the flux density along the $X$ axis produced by an optimized cylindrical EPM, 4 cylindrical EPMs, and 4 arc-shaped magnets. Operating distance $d$ of 240 mm.

![Flux Density Diagram](image)

**Figure 4.5** Comparison of $B_x$ produced by a single cylindrical EPM, structure $C_{1234}$ and structure $A_{1234}$. Operating distance $d$ of 30 mm.

With reference to the results in Fig. 4.5, the flux density generated by the array $A_{1234}$ is considerably higher than the one generated by the array $C_{1234}$ in some regions. For example, the maximum difference in the magnetic flux density between the two arrays is reached at $x= -24$ mm and the difference is 44 mT. This is a significant amount that would allow the actuation of small magnets if we consider that a magnitude of 103 mT is used in [12] to actuate two small magnets and magnetic flux densities between 4 mT to 14 mT have also been used to actuate small IPMs [54-56]. However, in our region of interest where the flux density is minimum (i.e., at $x=0$ mm), this difference
in magnetic flux density is reduced to 10 mT, as shown in Fig. 4.5. Due to this small difference, we believe that either an array of four cylinders or four ASMs can be used to improve the magnetic flux density at the center of the system and actuate a small IPM. Nevertheless, we choose to use ASMs for the following two reasons: i) in a real medical application, the IPM’s position is not restricted to the centre of the system, therefore higher magnetic torques can be imparted to it when it is moved away from the centre (for instance, for any other point along the X axis, 4 ASMs generate a higher magnetic flux density than 4 cylindrical EPMs); ii) the volume of the ring-shaped structure can be better filled if ASMs are used instead of cylindrical or cubic EPMs that would leave unused volume within the ring-shaped structure.

The analytical results show that $B_y$ is 0 mT along the X axis for the both types of arrays of magnets. For this reason, we do not show $B_y$ in any results. However, we present in Fig. 4.6 a 2D vector field representation of $|\mathbf{B}|$ created by 4 ASMs in plane $z=0$ (using Comsol).

Figure 4.6 shows that $|\mathbf{B}|$ approximates $B_x$ over a relatively large region around point P. Therefore, the $B_x$ component is mainly responsible for the transmitted torque on the IPM in Eq. 3.16.

Figure 4.6 Vector field of the magnetic flux density norm on the plane $z=0$ generated by the structure $A_{1234}$ when the operating distance $d$ is 30 mm. Scale on the right-hand side is given in Teslas.
4.2.4 Scaling up the magnetic systems and practical consideration

Although, in subsection 4.2.3, we carry out the analytical analysis and comparison of arrays of magnets with an operating distance \( d \) of 30 mm, we can scale up the EPMs to increase the operating distance. For instance, in our previous work [101], we use a more realistic operating distance of 240 mm which represents the scaled down distance multiplied by a factor of 8. Therefore, we can increase the dimensions of the EPMs presented in subsection 4.2.3 proportionally, by using the scaling factor of 8 as follows: for each arc-shaped magnet, we make \( L_2=30\times8=240 \text{ mm} \), \( \Delta \theta=\pi/6 \), \( r_2=50\times8=400 \text{ mm} \) and \( r_1=30\times8=240 \text{ mm} \) as shown in Fig. 4.7 (a). For each cylindrical EPM we make \( L_1=L_2 \), \( R=11.55\times8=92.4 \text{ mm} \) and its center is located at \( d_1=41.55\times8=332.4 \text{ mm} \) from the center of the system. An ASM with such dimensions could be customized by a manufacturer. However, it may be more practical to assemble cheaper and smaller arc-shaped permanent magnets to obtain the same results produced by a single custom ASM as shown in Fig. 4.7 (a) [109, 110].

The comparison of the \( B_x \) produced by these arrays of permanent magnets along the \( X \) axis is shown in Fig. 4.7 (b).

![Flux Density by Scaled up EPMs](image)

Figure 4.7 (a) Scaled up magnetic system (dimensions in mm). Left: a single custom arc-shaped permanent magnet; right: assembly with smaller arc-shaped magnets; (b) Comparison of \( B_x \) produced by the array of cylindrical magnets (denoted as \( C_{1234} \)) and the array of arc-shaped magnets (denoted as \( A_{1234} \)) when the operating distance \( d \) is 240 mm.
With reference to Fig. 4.7 (b), we find that the maximum difference in the magnetic flux density between the two arrays is reached at x = -192 mm (i.e. x = -24 mm * 8 or, in other words, this is 8 times the value found for the scaled down system) and the difference is 44 mT which is the same value found for the scaled down system. Figs. 4.5-4.6 and Fig. 4.7 (b) show that the flux density produced by the array of ASMs is always higher than 113 mT and it reaches its minimum value at the center of the system. Considering that the magnetic torque is proportional to the flux density as expressed in Eq. 3.16, we argue that, with the array A_{1234}, a minimum magnetic torque is exerted on the IPM when it is located at the center of the system. If the IPM is located at positions other than the center of the system, as will happen most of the time in the real application of DDS for WCE, a higher magnetic torque can be imparted to the IPM by the array A_{1234} than by using the array of cylindrical EPMs. This is due to the better use of the restricted space in the circle with a radius of d made by certain shapes such as arc-shaped permanent magnets when compared with cylindrical EPMs or cubic EPMs [110].

Since the center of the system is the critical point to obtain an improved flux density and, consequently, a useful magnetic torque, as it represents the longest distance to the EPMs, we conduct our experiments by placing the IPM at the center of the system. In regard to the orientation of the IPM in a DDS for WCE, it is expected that this will continuously change as the robotic capsule travels through the digestive system. Therefore, its magnetization vector will change direction and may affect the magnetic torque as predicted by Eq. 3.1. However, the assessment of changes in the magnetic torque due to variations in the IPM’s location and orientation to determine the limitations of the system are presented in Chapters 6-7.

It is also envisaged that a DDS for WCE will work simultaneously with additional modules such as an active locomotion system, an anchoring mechanism, and a localization and orientation module. All these modules must be compatible with the magnetic DDS. The active locomotion system would allow the physician to take the robotic capsule to the region of interest by controlling the capsule’s position remotely. For example, [76] presents a torque-driven magnetic system for active locomotion that may be compatible with our proposed magnetic DDS. Once the WCE reaches the
target area, the anchoring mechanism would allow the physician to stabilize the robotic capsule before releasing the drug compounds. In this way, the robotic capsule will be able to resist the peristaltic force in the gastrointestinal tract, the magnetic force generated by the EPMs and the reaction forces generated within the robotic capsule while the drug is being released. A force-driven magnetic system that allows the capsule to anchor is presented in [10] and it can be compatible with our proposed DDS. Finally, a localization and orientation module embedded in the robotic capsule, such as the localization system based on positron emission markers presented in [73] which is compatible with our magnetic DDS, would provide information to adjust the EPMs’ position and orientation and compensate for misalignments with the IPM if needed. The incorporation of all these additional modules in the restricted volume of a WCE emphasizes again the necessity of miniaturizing the IPM and optimizing the external magnetic system to achieve efficient magnetic linkage at longer operating distances.

For practical reasons, we have decided to experiment with the scaled down magnetic system made of an array of arc-shaped permanent magnets (i.e., structure $A_{1234}$) and the details are presented in the next two sections.

### 4.3 Experimental setup for magnetic interactions

The general coordinate system XYZ, defined in Fig. 3.1, is associated with the fabricated plastic case shown in Fig. 4.8 (a) that possesses $30^0$ angle indicators and allows the manual rotation of the array of ASMs.
Figure 4.8 (a) EPMs fixed on the aluminum case and rotated by $\theta_{EPM} = 30^\circ$ degrees, (b) Experimental setup consisting of the measurement instruments and the array of arc-shaped permanent magnets.

The angle $\theta_{EPM}$ represents the misalignment angle between the X and $X_a$ axes as shown in Fig. 3.3. Figure 4.8 (a) shows, for example, the EPMs rotated by an angle $\theta_{EPM}$ of $30^\circ$. In the experimental analysis of the magnetic flux density created by the external magnetic system, we align $X_a$ and X axes (i.e., $\theta_{EPM} = 0^\circ$) because there is no need to rotate the external magnetic system. However, in the experimental analysis of the magnetic torque imparted to the IPM, $\theta_{EPM}$ takes values from $0^\circ$ to $360^\circ$ allowing the manual rotation of the external magnetic system.

A 3-channel Gauss meter (Lakeshore-Model 460) was used to measure the magnetic flux density generated by the ASMs whose dimensions and magnetization grade are defined in Section 4.2.3. A torque gauge (HTG2-40 supplied by IMADA) with its respective torque sensor held the IPM at the center of the system. The torque sensor and the probe tip of the Gauss meter were mounted on plastic holders which were also fabricated using a 3D printer. Both the torque sensor and the probe tip of the Gauss meter can be moved along the X and Z axes and the arrays of magnets can only be moved along the Y axis. These displacements were controlled by a micromanipulation system based on an X-Y-Z stage, as shown in Fig. 4.8 (b). The experimental setup shown in Fig. 4.8 (b) was used to validate the theoretical results for $B_x$ generated by the array of ASMs and the transmitted magnetic torque on an IPM, as presented in subsections 4.4.1-4.4.2.

4.4 Experimental results with a prototype of DDS

In the first series of experiments, we measured the magnetic flux density $B_x$ produced by the individual segments $A_1$ and $A_3$. In the subsequent experiments, we experimentally evaluated the effect of having multiple segments, thus we measured $B_x$ produced by $A_1$ and $A_2$ acting simultaneously, which we designate as $A_{12}$. Similarly, we measured $B_x$ produced by $A_{34}$ and $A_{1234}$ which was the contribution of all the segments acting simultaneously. In all these experiments, the magnets were fixed in their respective positions (i.e., the Y axis did not move). The z-position of the probe tip was adjusted until it reached $z=0$ and then the probe tip was moved from -21 mm to
24 mm along the X axis. All these experimental results were compared with the analytical results and are presented in the following subsection.

### 4.4.1 Experimental results for the magnetic flux density

We can see that the magnetic flux density is enhanced when multiple magnets are added in the system. The maximum $B_x$ measured at the center of the system is 114.4 mT with an array of four arc-shaped permanent magnets as shown in Fig. 4.9 (c).

Figure 4.9 $B_x$ produced by arc-shaped magnets: (a) radially magnetized ($A_1$ and $A_{12}$), (b) tangentially magnetized ($A_3$ and $A_{34}$), and (c) the array $A_{1234}$.

### 4.4.2 Experimental results for magnetic torques

In the second series of experiments, we were interested in measuring the magnetic torque $\tau_z$ exerted on the 6.35 mm cubic IPM with the magnetization of 1.25 Tesla (N40) only by segment $A_1$. These dimensions and magnetization grade of the IPM are specified in subsection 4.4.2, but these parameters are varied, in subsection 4.4.3, to determine the smallest IPM to be embedded in the robotic capsule. Even though, for the same volume and magnetization grade, cylindrical IPMs can produce higher
magnetic torques than cubic IPMs [102], we decided to conduct our experiments with the worst scenario (i.e., with cubic IPMs). In the subsequent experiments, we verified the effect of having multiple segments, thus we measured $\tau_z$ produced by $A_{12}$, $A_{123}$ and $A_{1234}$. In all these experiments, the IPM is fixed at the center of the system (i.e., $X=Y=Z=0$) and its magnetization vector $m$ was aligned with the X axis at all times. The ASMs rotated about the Z axis with increments of $30^0$ and therefore $\gamma$ in Eq. 3.16 always equals $\theta_{EPM}$. The comparison between the analytical results, which are estimated using Eq. 3.16, and the experimental results for the transmitted magnetic torque $\tau_z$ is presented in Fig. 4.10.

![Graph](image)

Figure 4.10 $\tau_z$ produced by single and multiple permanent magnets on the cubic IPM.

We can see that the combination of multiple magnets not only improves the magnetic field at the center of the system but also the peak torque on the IPM. The maximum torque exerted by the array $A_{1234}$ on the cubic IPM was measured as 26 mNm. Although the assessment of the magnetic torque for different IPM’s positions and orientations is not within the scope of this chapter, we do present some experimental results in [102] for an IPM which is not located at the center of the system. We use the array $A_{1234}$ as the source of the rotating magnetic field to actuate a slider-crank mechanism and the details of this mechanism and the experimental results are presented in subsection 4.4.3.
4.4.3 Experimental results for piston force

We connected the cubic IPM to a slider-crank mechanism to convert the rotational motion of the IPM into the translational motion of a piston. The IPM is inserted in its case and its magnetization vector, \( m \), is always parallel to the vector that is projected on the plane \( z=0 \) and the tail and tip of which are located at the center of the crankshaft and the center of the crankpin, respectively, as shown in Fig. 3.34 (a).

The IPM case can house cubic magnets ranging from 3.175 to 10 mm. Fig. 3.34 (b) shows the components of the slider-crank mechanism (i.e., the grey disk, the connecting rod (B) and the green piston (C) shown in Fig. 3.34 (b)). The IPM is inserted into the IPM case which is held by the yellow platform (A) that is fixed and attached to the platform support (G). The IPM freely rotates about the crankshaft the center of which is aligned with the center of the external magnetic system.

It should be noted that since the IPM is physically connected to the crankshaft, the angle of \( m \) with respect to the X axis equals the crankshaft angle \( \alpha \) defined in Fig. 3.15. \( \beta \) which is the angle formed by the external magnetic system and the X axis. This angle ranges between \(-180^0\) and \(180^0\) (see Fig. 3.34 (b)). In order to measure the force \( P \) delivered to the piston (defined in Section 3.4) when the IPM rotates, we used a helical spring that is compressed as the piston moves forward and creates the spring force \( F_s \). The spring is extended when the piston is moved back to its original position, as shown in Fig. 4.11.

![Figure 4.11 Components of the slider-crank mechanism and the mechanical spring to measure the piston force.](Image)
The slider-crank mechanism was fabricated with the length of the crank R=3 mm and the length of the connecting rod L=9 mm which are dimensions compatible with the size of a commercial WCE. Using these dimensions and Eq. 3.8, we obtain the minimum and maximum piston’s positions as \( x_{\text{min}} = 6 \text{ mm} \) and \( x_{\text{max}} = 12 \text{ mm} \), respectively.

In our experiments, we manually rotated the ASMs and the cubic IPM rotated at the same time, compressing the spring when \( \beta \) changed from 180° to 0° and extending it when \( \beta \) changed from 0° to -180°. A laser (optoNCDT 1700 by Micro-Epsilon), as shown in Fig. 4.12 (b), was used to measure the stroke \( x \) of the piston expressed by Eq. 3.21. The beam of the laser was targeted on the reflective surface that was connected to the piston, as shown in Fig. 3.34 (b). The laser reading was used to estimate the crank angle \( \alpha \) and the spring force \( F_s \) which are both dependent on the stroke \( x \), as expressed by Eq. 3.10 and Eq. 3.21, respectively. Once \( F_s \) and \( \alpha \) are estimated, we use Eq. 3.19 to estimate the crank-shaft torque \( \tau_c \) needed to balance the force \( F_s \) (Note: the piston force \( P = F_s \) when using Eq. 3.19).

Figure 4.12 (a) shows the entire system at the initial position. At this position, the magnetization vectors of the segment \( A_2 \) and the cubic IPM are pointing towards \( \beta=180^\circ \) and the position of the wrist pin (i.e, point B shown in Fig. 3.15) is 6 mm away from the center of the crank.
Figure 4.12 (a) External magnetic system powering the slider-crank mechanism and rotated by $\beta=180^\circ$, (b) the laser was used to measure the piston displacement along the X axis.

We conducted experiments with a variety of cubic IPMs to assess the capability of the system to convert the magnetic torque into a piston force. Table 4.1 shows the specifications of different IPMs and Fig. 4.13 shows the spring force $F_s$ which equals the magnitude of the piston force $P$ but its direction is opposite to the piston force direction.

Table 4.1 Specifications of IPMs used in the experiments

| Magnetization grade | $|m| [T]$ | Size [mm] |
|---------------------|----------|-----------|
| N50                 | 1.40     | 3.175     |
| N50                 | 1.40     | 4         |
| N40                 | 1.25     | 5         |
| N40                 | 1.25     | 6.35      |
| N40                 | 1.25     | 10        |
Figure 4.13 Piston force response with a variety of cubic IPMs ($P=F_s$). It shows the compression and extension of the spring in the entire cycle.

Figure 4.13 shows that, for IPMs smaller than 6.35 mm, the spring reaches a maximum compression at which the piston exerts its peak force. Although peak forces are not required to release the drug from the reservoir, once the piston force reaches its peak value, the piston will not move forward beyond this point. For instance, for the smallest IPM (3.175 mm), the peak force is obtained when the EPMs are rotated by $\beta=30^0$, while this peak is reached at $\beta=0^0$ for IPMs of 4 mm and 5 mm. At the point when the IPMs cannot further compress the spring, the EPMs provide the maximum magnetic torque. However, if we continue rotating the ASMs until they reach approximately $\beta=-90^0$ for IPMs of 4 and 5 mm, the spring is extended (i.e. released) abruptly.

Figure 4.14 shows the crankshaft torque $\tau_c$ estimated using Eq. 3.10, Eq. 3.19 and Eq. 3.21. For instance, when the EPMs were manually rotated until they reached $\beta=60^0$, the piston’s position $x$ was measured as 7.05 mm for the smallest IPM (3.175 mm). The force on the piston is estimated to be $F_s = K\Delta x = 1.59 \times (7.05 - 6) = 1.67$ N as shown in Fig. 4.13. Since only position $x$ is measured with the laser sensor, we can estimate the crank angle $\alpha$ by using Eq. 3.10 and we found $\alpha$ to be 121.81°. We then used $\alpha$ and force $F_s$ in Eq. 3.21 to estimate crankshaft torque $\tau_c$ of 3.51 mNm as shown in Fig. 4.14.
Figure 4.14 Crankshaft torque response with a variety of cube IPMs. It shows the compression and extension of the spring in the entire cycle.

This torque $\tau_c$ exerted by the spring on the crankshaft should be equal in magnitude (but in the opposite direction) to the magnetic torque $\tau_z$ produced by the EPMs acting on the IPM. Thus, we can use Eq. 3.16 to validate the result obtained for the crankshaft torque. When the EPMs are rotated by $\beta=60^\circ$, the crank angle $\alpha$ was calculated as $121.81^\circ$ with respect to the X axis. Since the IPM is physically connected to the crankshaft, the magnetization vector $\mathbf{m}$ forms an angle of $121.81^\circ$ with respect to the X axis, as shown in Fig. 4.15.

![Figure 4.15 Vector representation when the EPMs are oriented at $\beta=60^\circ$. This vector representation is a top view of the coordinate system defined in Fig. 3.34 (b).](image)

We use the following values in Eq. 3.16: $V = 3.2 \times 10^{-8}$ m$^3$ (for the smallest IPM), magnetization of 1.4 T, $|\mathbf{B}|$ was measured as 114.4 mT at the center of the system (see Fig. 4.9 (c)) and $\gamma = 60^\circ - 121.81^\circ = -61.81^\circ$ and we estimate $\tau_z$ to be -3.59 mNm. The negative value of the magnetic torque $\tau_z$ indicates that this torque is in the clockwise direction, thus it approximately balances the crankshaft torque $\tau_c$ of 3.51 mNm in the counter clockwise direction shown in Fig. 3.15.

There is only a small difference between the magnitudes of $\tau_z$ and $\tau_c$ (less than 0.1 mNm), thus validating our results. We postulate that this small difference could be due to imperfections in the experimental setup such as friction force and clearances at the joints. In order to overcome the limitations associated with the accuracy of 3D printing, all the components of the slider-crank mechanism can be fabricated more precisely using, for example, the LiGA process [11]. The improvement in the fabrication of these
components will be useful for the final integration of the mechanism in commercial WCEs.

**4.4.4 Drug delivery capability**

We fabricated a cubic piston with a cross-sectional area of 105.6 mm$^2$ (12 mm x 8.8 mm) and a maximum stroke of 6 mm ($X_{\text{max}}-X_{\text{min}}=2R=6$ mm). These dimensions give a total drug reservoir volume of 0.633 mL. Compared to the reservoir of the capsule-based drug delivery systems reported in the literature [1, 23], which ranges between 0.17 mL and 1 mL, this is a reasonable drug reservoir volume. We can fabricate the slider-crank mechanism with a longer stroke or a larger cross-sectional area to easily bring the drug reservoir volume to 1 mL. For example, if we increase the length of the crank to $R=5$ mm and use a cylindrical piston with a cross-sectional area of 95 mm$^2$ (i.e., by considering the typical diameter of a WCE which is 11 mm [66]), then we obtain a maximum stroke of $2R=10$ mm. These dimensions would result in a total drug reservoir of 0.950 mL.

The number of doses that our fully controllable prototype of DDS can release depends on the pharmaceutical or treatment needs. For instance, if the total drug reservoir volume is divided by 6, with volumes of 0.105 mL each, then the maximum number of doses to deliver would be 6. In this case, we can rotate the external magnetic system to make the piston advance by increments of 1 mm each time. The first increment can be obtained when the crank angle changes from $180^0$ to $123^0$, the second drug release requires the crank angle to change from $123^0$ to approximately $100^0$. The third release would be possible by decreasing the crank angle to $80^0$ and we can continue releasing the drug until the crank angle becomes $0^0$ in a nonlinear fashion.

If more than 6 doses are required, the number of doses and release amount can be precisely controlled by making the piston advance in smaller increments as long as the torque load of the drug payload remains under the peak torque value imparted to the IPM. Finally, the release rate will depend on the rotational speed of the external magnetic system. Although, in this chapter, we manually rotated the external magnetic system, its rotational speed could be more precisely controlled by using motors and a control station along with a joystick, as illustrated in Section 3.2.
4.5 Conclusions

It is highly desirable to include an effective and accurate DDS in the next generation of WCE. Several requirements must be fulfilled, however, for the successful development of such a system, and these include the active actuation of an untethered releasing mechanism that allows the control of variables such as the release rate, release amount and number of doses. In this chapter, the focus has been on the design and shape optimization of an external magnetic system and the dimensions of the IPM to remotely actuate a drug release mechanism for CE.

We investigate the most suitable external magnetic system to produce the highest rotating magnetic field under which a small internal magnet (i.e., IPM) could be used. We compare a single cylindrical permanent magnet against arrays of cylindrical and arc-shaped permanent magnets (ASMs) as the source of the rotating external magnetic field. We find that, for the same volume, the arrays of permanent magnets can produce stronger magnetic fields than a single cylindrical magnet. We also find that either cylindrical or ASMs are appropriate to improve the magnitude of the magnetic field at the center of the system. However, ASMs can produce higher magnetic fields in regions where the IPM is also expected to be. Therefore, ASMs provide advantages over the cylindrical magnets in reducing the volume and weight of the external magnetic system. Since these advantages are important to reduce the fabrication costs and also ease the maneuverability of the external magnetic system, we used ASMs in our experiments and verified that the combination of four ASMs (at optimized locations and orientations) not only improved the magnetic field at the center of the system but also the peak torque on the IPMs. Since these results are based on analytical models that are valid for different sizes of magnets, we conclude that an array of multiple ASMs can be scaled up and placed at longer distances from the center of the system to actuate a small IPM embedded in a robotic capsule.

The magnetic flux density generated by four ASMs was measured at the center of the system as 114 mT. Several cubic IPMs acting independently were used to actuate the piston that would expel drug out of a reservoir. We assessed the capability of each cubic IPM to convert the magnetic torque into a piston force, and found that the smallest cubic IPM (i.e., 3.175 mm) produced a peak piston force of 1.67 N.
Considering that a peak piston force of only 820 mN is needed to release a variety of drug compounds [101], we conclude that even the 3.175 mm cubic IPM is sufficient to release drugs and further miniaturization of the IPM is still possible. Nevertheless, a further miniaturization and compactness of the slider-crank mechanism is also needed to leave sufficient room within the capsule robot to integrate additional modules such as the image guidance and anchoring mechanism to improve the accuracy of the drug release procedure. The 3.175 mm cubic IPM which we used is the smallest size that has been used in a prototype of the robotic capsule (if compared to the ones reported in the literature [9, 10, 12, 63, 65, 75, 76]). Therefore, our optimized external magnetic system guarantees that an adequate amount of magnetic field is produced to actuate the IPM while providing the following benefits: a longer operating distance, enough volume for the drug reservoir, high control over the number of doses and the release amount. Furthermore, the optimized magnetic system is able to actuate the drug release module when the capsule is located not only at the centre of the system, but also at any other point within the region of operation, which is of great advantage for the irregular transport process of the capsule through the biological tract.

In regard to the experimental results for the crankshaft torque, we have found that a peak torque of about 3.5 mNm (which is converted into a peak piston force of 1.67 N) is adequate to actuate the piston. In order to generate smooth movements in the piston, however, the magnetic system should be designed in such a way that it is always able to generate a magnetic peak torque that exceeds the crankshaft torque requirement at any angle of orientation of the EPMs. However, the peak force and torque are not always required to release the drug compound. Since ASMs perform better than cylindrical magnets, the next chapter (Chapter 5) focuses on the optimization (optimal angular positions and size optimization) of multiple ASMs. As we have briefly proposed in subsection 3.2.1, the external magnetic system can be powered by electric motors. More specifically, the scaled up EPMs could be mounted on a 6-DOF platform powered by the motors and controlled via a joystick. To ease the demand on the motors and allow maneuverability of the external magnetic system, it is desirable to minimize the volume and weight of the EPMS. These are our main motivations to optimize the dimensions and angular positions of the EPMs which are carried out in Chapter 5.
Chapter 5
Angular Position and Size Optimization of the External Magnetic System

The main objectives of this chapter (Chapter 5) are to optimize the angular position and size of the ASMs to generate sufficient magnetic flux density and torques to actuate a DDS for WCE while minimizing the total weight of the ASMs. The theoretical analyses carried out in Chapter 5 are valid for any dimensions of the ASMs and further details of the scaling laws have been provided in Chapter 6 (more specifically in Section 6.1.1). Nevertheless, we have decided in Chapter 5 to conduct theoretical and experimental analyses with a scaled down magnetic system only for practical reasons.

5.1 Angular position optimization

We have carried out design and shape optimization of the external magnetic system [101, 111] and also shape optimization of the IPM [102]. We have fabricated a prototype of the external magnetic system with four ASMs and its results reported in Chapter 4 show that sufficient piston force is generated to expel drug out of a chamber. In this chapter, we continue on the optimization of the external magnetic system to further enhance the magnetic field. Specifically, we present in Section 5.1 how to increase the external magnetic flux density by finding optimal angular positions for 12 off-the-shelf ASMs. The improvement in the magnitude of the magnetic field produces a higher magnetic torque on an IPM that is to be embedded in a prototype of CE.

The analysis presented in Section 5.1 can be used for any number of segments, but for practical reasons, we use three segments of each type of ASM shown in Fig. 4.2 (c). With reference to Fig. 4.2 (b), the dimensions of each ASM are: $r_1=30$ mm, $r_2=50$ mm, length of 30 mm, angular width $\Delta \theta = \pi/6$, and all have the same magnetization grade of 1.32 T (i.e., N45) (these are the same specifications described in Section 4.2.3). A 3.1 mm cubic IPM with the magnetization grade of 1.4 T (i.e., N50) is placed in the prototype of a capsule. With 12 ASMs, there are different possible configurations to place in a ring-shaped structure. We are interested in finding the optimal configuration (i.e., optimal angular position for each ASM within the ring-shaped structure) to transmit the highest possible torque on the IPM.
5.1.1 Theoretical methods

We use the analytical model for the magnetic field created by ASMs [106] which is based on the Coulombian model. We are interested in maximizing $|B| = \sqrt{B_x^2 + B_y^2}$ at the location of the IPM (note that $B_z$ does not contribute to $\tau_z$). Thus, we need to understand the variation of the flux densities $B_x$ and $B_y$ at the center of the system produced by the radially and tangentially magnetized segments $A_i$, $i=1,2,3,4$ when they follow a circular trajectory of radius $r=40$ mm ($r = \frac{r_1 + r_2}{2}$) as shown in Fig. 5.1.

Figure 5.1 Segment $A_3$ follows a circular trajectory of radius $r=40$ mm.

Figure 5.2 $B_x$ at the centre of the system generated by $A_3$ as it moves along the circular trajectory of radius $r=40$ mm.
Figure 5.2 shows the variation of $B_x$ at the centre of the system produced by segment $A_3$. $B_x$ was maximum at $\theta_p=90^0$ and its maximum magnitude at this position was $B_{x_{\text{max}}}=19.1 \text{ mT}$. In other words, the optimal angular location of the segment $A_3$ along the circular trajectory is found when we place it at $\theta_p=90^0$. We denote this as $A_3^{90^0}$. If we placed $A_3$ at $\theta_p=60^0 (A_3^{60^0})$, we would obtain $B_x=16.5 \text{ mT}$ at the centre of the system.

Similarly, $A_3^{120^0}$ produces $B_x=16.5 \text{ mT}$ at the centre of the system (see Fig. 5.2). We also aim to find the maximum contribution to the magnetic flux density $B_x$ at the centre of the system that the arc-shaped magnet $A_1$ can generate when it follows the same circular trajectory. The results are shown in Fig. 5.3. $B_x$ is maximum at $\theta_p=180^0$ and its maximum magnitude at this angular position is $B_{x_{\text{max}}}=37.5 \text{ mT}$. In other words, the optimal position of the segment $A_1$ in the circular trajectory is found when we place it at $\theta_p=180^0$ (i.e., $A_1^{180^0}$).

![Graph showing $B_x$ at the centre of the system by $A_1$](image)

Figure 5.3. $B_x$ at the centre of the system generated by $A_1$ when it follows the circular trajectory of radius $r=40 \text{ mm}$.

If we placed $A_1$ at $\theta_p=120^0$ (i.e., $A_1^{120^0}$), we would obtain $B_x=18.8 \text{ mT}$ at the centre of the system. Similarly, $A_1^{150^0}$ produces $B_x=32.5 \text{ mT}$ at the centre of the system (see Fig. 5.3). Table 5.1 compares the contributions to $B_x$ at the centre of the system by each segment $A_1$ and $A_3$ as a function of $\theta_p$ between $90^0$ and $180^0$, which is the range of interest since within this range, we find the positive peak values of $B_x$ as shown in
Figs. 5.2-5.3. Similarly, $B_y$ produced by both segments at the centre of the system is presented in Fig. 5.4.

![Diagram showing $B_y$ at the centre of the system]

Figure 5.4 $B_y$ at the centre of the system generated by $A_1$ and $A_3$ along the circular trajectory of radius $r=40$ mm.

<table>
<thead>
<tr>
<th>$\theta_p$ [deg]</th>
<th>$A_1$ [mT]</th>
<th>$A_3$ [mT]</th>
</tr>
</thead>
<tbody>
<tr>
<td>90</td>
<td>0</td>
<td>19.1</td>
</tr>
<tr>
<td>120</td>
<td>18.8</td>
<td>16.5</td>
</tr>
<tr>
<td>150</td>
<td>32.5</td>
<td>9.5</td>
</tr>
<tr>
<td>180</td>
<td>37.5</td>
<td>0</td>
</tr>
</tbody>
</table>

Table 5.1 Contributions to $B_x$ at the centre of the system by $A_1$ and $A_3$

We can use Table 5.1 to find the optimal configuration for the array of magnets that enables us to obtain the maximum $B_x$ at the centre of the system. For instance, at $\theta_p=90^0$, $A_3$ contributes to $B_x$ with 19.1 mT while the contribution of $A_1$ at this position is nil. Similarly, at $\theta_p=120^0$, $A_3$ contributes to $B_x$ with 16.5 mT while the contribution of $A_1$ at this position is slightly higher. Following this methodology, we find that the optimal configuration to accommodate magnets $A_1$ and $A_3$ in the region of interest (i.e., $90^0 < \theta_p < 180^0$) is as shown in Fig. 5.5. By considering the symmetry of the magnetic system with respect to the X and Y axes, we find the optimal configuration in the entire circular trajectory to be the one shown in Fig. 5.6 that consists of 10 segments radially...
magnetized (5 segments of the type A1 and 5 segments of the type A2) and two segments tangentially magnetized. By following the same procedure with the variation of B_y, we find the same results. We can also use a curve fitting process and obtain the same optimal configuration if we note that the variation of B_x and B_y can be expressed as sinusoidal functions with the peak amplitudes of 19.1 mT and 37.5 mT.

Figure 5.5 Optimal configuration with 4 segments in the region 90° < θ_p < 180°.

Figure 5.6 Configuration1: optimal configuration with 12 segments in the region 0° < θ_p < 360°.

In this optimal configuration, called Configuration1, the contribution of each segment gives a total B_x = 318.5 mT at the centre of the system and B_y = 0 mT. Considering that we aim to maximize the magnetic field with 3 segments of each type rather than 10 segments with radial magnetization plus two segments tangentially magnetized, we can replace A^{120°}_1 by A^{120°}_3 and replace A^{60°}_2 by A^{60°}_3 and also taking into account the symmetry of the system, we find the second optimal configuration, called Configuration2, as shown in Fig. 5.7. In this configuration, we obtain B_x = 309.5 mT at the centre of the system which is slightly less than B_x obtained in Configuration1 (and B_y = 0 mT for Configuration2).
Figure 5.7 Configuration2: second optimal configuration with 12 segments in the region $0^0 < \theta_p < 360^0$.

Two other optimal configurations are presented in Figs. 5.8-5.9. For example, the best possible configuration in the region $0^0 < \theta_p < 180^0$ is generated in Configuration3 while the best possible arrangement with 4 ASMs in the region $0^0 < \theta_p < 360^0$ is given by Configuration4. We have proposed the latter in [101] but with cylindrical magnets.

Figure 5.8 Configuration3: 7 segments placed in the region $0^0 < \theta_p < 180^0$ in the best possible configuration.

Figure 5.9 Configuration4: 4 segments placed in the region $0^0 < \theta_p < 360^0$ in the best possible configuration.

In Fig. 5.10, we present the variation of $B_x$ along the X axis for the 4 previous optimal configurations. The flux density $B_y$ is 0 mT along the X axis in all these configurations, except in Configuration3 where $B_y$ is 0 mT only at the centre of the system. However, since the IPM will be located at the centre of the system, therefore $|B| = B_x$.

Figure 5.10 $B_x$ created by four different configurations.
By using Eq. 3.16, we estimate a peak magnetic torque of $\tau_z = 11.35$ mNm at $\gamma = 90^0$ (i.e., when the misalignment angle between $\mathbf{B}$ and $\mathbf{m}$ is 90$^0$) if the IPM is actuated by Configuration1. On the other hand, if the IPM is actuated by Configuration2, the magnitude of $|\mathbf{B}|$ decreases from 318.5 to 309.5 mT and thus, the peak magnetic torque $\tau_z$ is estimated to be 11.02 mNm, which is approximately the same as the one obtained with Configuration1. In the next section, we compare these analytical results against the experimental results.

5.1.2 Experimental methods

Similar to the experimental system used in Chapter 4, a 3-channel gauss meter (Lakeshore-Model 460) was used to measure the magnetic flux density. A torque gauge (HTG2-40 made by IMADA) with its respective torque sensor held the IPM at the centre of the system. Both the torque sensor and the probe tip of the gauss meter can be moved along the X and Z axes and the arrays of magnets can only be moved along the Y axis. These displacements are controlled by a micromanipulation system constructed of XYZ stages as shown in Fig. 5.11. Fig. 5.11 also depicts the optimal magnetic structure called Configuration2 which was also used in subsection 3.6.3.1.
magnets.

5.1.2.1 Magnetic flux density

The comparison between analytical model and experimental results for the $B_x$ produced by radially and tangentially magnetized ASMs is presented in Figs. 5.12-5.13.

Figure 5.12 $B_x$ at the centre of the system produced by segment $A_3$ that moves along a circular trajectory with radius of 40 mm.

Figure 5.13 $B_x$ at the centre of the system produced by segment $A_1$ that moves along a circular trajectory with radius of 40 mm.
The comparison between the analytical model and experimental results for the total $B_x$ generated by Configurations 2, 3 and 4 is presented in Figs. 5.14-5.16.

Figure 5.14 $B_x$ along the X axis generated by Configuration 2.

Figure 5.15 $B_x$ along the X axis generated by Configuration 3.

Figure 5.16 $B_x$ along the X axis generated by Configuration 4.

These experimental results for $B_x$ generated by Configurations 2, 3 and 4 validate the analytical models that we use for the optimization of the magnetic system. However,
we found a small difference in Fig. 5.14 between the theoretical and experimental results. For instance, at the centre of the system, we obtain a theoretical value of $B_x=309.5$ mT, but in our experiment, we measured $B_x=303.2$ mT because the Configuration2 was created by placing the ASMs in a circular trajectory of radius $r=40.4$ mm instead of $r=40$ mm. Due to this small difference in the radial distance, we obtained a slightly lower magnetic field at the centre of the system. Since Configuration2 is the optimal angular configuration for 12 off-the-shelf ASMs and produces the highest magnetic field, we used it as the source of the external magnetic field to actuate the piston as it is presented in Section 5.1.2.3.

### 5.1.2.2 Magnetic torque

The comparison between the analytical model and experimental results for $\tau_z$ exerted on the IPM by the external magnetic system produced by configurations 2 and 3 is presented in Figs. 5.17-5.18. $\theta_{EPM}$ is the angle by which the external magnetic system is rotated (see Fig. 3.1).

![Graph](image)

Figure 5.17 Magnetic torque $\tau_z$ exerted on the 3.1 mm cubic magnet by the external magnets set in Configuration2.

The peak torque of 10 mNm on the 3.1 mm cubic IPM was possible when the external magnetic field was created by the array of magnets set in Configuration2 (see Fig. 5.17). For this reason, we used the Configuration2 to actuate the drug release mechanism. Its details are presented in subsection 5.1.2.3.
5.1.2.3 The slider-crank mechanism

We tested the capability of the slider-crank mechanism to convert the magnetic torque $\tau_z$ into a piston force $P$. Therefore, we inserted the 3.1 mm cubic IPM into its case that is connected to the slider-crank mechanism shown in Fig. 3.33. We used a helical spring and followed the same method described in subsection 3.6.3.2 and subsection 4.4.3 to measure the piston force and crank-shaft torque. The experimental setup shown in Fig. 3.34 was used.
Figure 5.19 The experimental piston force $P$ under the external magnetic field created by Configuration 2.

Figure 5.19 shows the piston force generated as the external magnetic system rotates one full cycle, compressing the helical spring in the left hand side of the curve and extending it on the other half of the curve. The experimental peak force of 3.6 N was obtained at the maximum compression of the spring, when $\beta = 0^\circ$. At this point, we estimated with the laser reading $\alpha = 94^\circ$ and the misalignment angle between the IPM magnetization direction and the direction of the external magnetic field was $94^\circ$, producing the peak torque of 10.7 mNm shown in Fig. 5.20. This peak torque on the crankshaft estimated with the laser reading is the same magnetic peak torque exerted on the IPM and measured with the torque gauge shown in Fig. 5.17, thus it validates the theoretical results.

![Crankshaft Torque](image)

Figure 5.20 Crankshaft torque $\tau_c$ generated under the external magnetic field created by Configuration 2.

### 5.1.3 Results and discussion

We have presented how to enhance the external magnetic flux density by finding an optimal configuration with 12 off-the-shelf arc-shaped permanent magnets. This optimization was carried out by using analytical models to describe the magnetic field created by ASMs with different magnetization directions. Our experimental results for the magnetic field verified the accuracy of the analytical model used for the
optimization process. We used the optimized configuration, called Configuration2, as the source of the magnetic field to make a 3.1 mm cubic magnet (i.e., the IPM) rotate about its own axial axis.

The maximum magnetic flux density produced by Configuration2 at the centre of the system was measured as 303 mT. Under this flux density, a peak magnetic torque of about 10 mNm was exerted on the 3.1 mm cubic IPM. This torque enables the piston to push with a peak force of 3.6 N. Considering that a peak piston force of only 820 mN is needed to release a variety of drug compounds [101], we conclude that our prototype of external magnetic system can be scaled up and the operating distance can be increased to allow the ASMs to surround a patient’s body. Since higher magnetic torques are imparted to the same IPM as we continue enhancing the magnetic flux density, we present the size optimization of the ASMs in Section 5.2 to further increase the magnetic flux density while simultaneously using the minimum possible volume in the external magnetic system.

5.2 Size Optimization with single ASMs

Magnetic coupling systems are used in many applications due to providing i) physical isolation between the driver magnetic source and the driven load, ii) no requirements for lubrication, and (iii) non-destructive torque overload [112]. The use of different magnetic coupling forms for actuation systems in biomedical applications have become an important area of research because such magnetic systems present no harm to living tissues [12, 113].

Magnetic forces and torques are commonly used in the actuation of a variety of mechanisms. For instance, in a tetherless robotic intervention presented in [114], an electromagnet (i.e., an MRI machine) is used as the driving magnetic system that exerts a magnetic force on a driven load that is connected to a needle. Similarly, magnetic coupling has been used to remotely actuate different mechanisms embedded in prototypes of robotic capsules to enhance the existing WCE’s capability as a complementary diagnostic medical tool [97]. This may allow medical practitioners to perform more complex procedures such as biopsy [12, 63], wireless insufflation [65], and active locomotion [115].
However, if these prototypes are to be implemented in a more realistic environment, larger operating distances between the driver and the driven magnets and further miniaturization of the driven magnets are required [101]. Therefore, the optimization of both the driver and driven magnets, which has been neglected in the research on capsule robots, is important to overcome these two limitations. We have presented the design and shape optimization of a driving magnetic system and also the optimization of the angular position of multiple ASMs, in Chapter 4 and Section 5.1, respectively. Furthermore, the shape optimization of the IPM has been presented in Section 3.6. These optimization processes have helped us to obtain an efficient magnetic linkage (i.e., an enhanced magnetic torque imparted to the IPM) while overcoming the aforementioned limitations.

Nonetheless, the primary aims of this section (Section 5.2) are: the size optimization of ASMs (i.e., thickness, angular width and length) and determining the effect on the magnetic linkage due to changes in the dimensions of ASMs. These are carried out by using analytical solutions which allow fast global optimization and are more efficient and capable of facilitating physical understanding, than the time consuming finite-element methods [84, 116]. Additionally, we use a statistical analysis (i.e., ANOVA) to determine the order of priority in which the dimensions of the ASMs should be changed to obtain efficient magnetic linkages. Although we present, in this section, the size optimization of a driving magnetic system to specifically actuate a drug delivery system (DDS) for WCE, the results and conclusions can be also applied to actuate different on-board mechanisms in magnetic capsule robots.

5.2.1 External magnetic system

The ring-shaped external magnetic system (i.e., the driving magnetic system) can be made of radially magnetized ASMs, tangentially magnetized ASMs or a combination of them as shown in Fig. 5.21. In this section, we present the size optimization of both types of ASMs working independently. However, the size optimization of the combination of these two types acting simultaneously is presented in Section 5.3.
Figure 5.21 (a) The external magnetic system made of arc-shaped permanent magnets $A_i$, i=1,2,3,4. Parameters of an arc-shaped permanent magnet: (b) a radially magnetized segment and (c) a tangentially magnetized segment.

The driving magnetic system, the IPM and the slider-crank mechanism are the main components of this drug release mechanism (see Fig. 3.1). We aim to optimize the dimensions of the ASMs (i.e., thickness: $\Delta r$, angular width: $\Delta \theta$, and length: $\Delta z$) to obtain an optimized magnetic field at the centre of the system where the IPM is located and subsequently obtain an optimized magnetic torque driving the drug delivery mechanism.

The centre of the system, called point P in Fig. 5.21 (a), is located at the centre of a circle with a radius of $r_1$. The thickness of each ASM $\Delta r$ is given by the difference between their external and internal radii, $r_2$ and $r_1$, respectively as shown in Figs. 5.21 (b)-(c). The length of each segment is $\Delta z = z_2 - z_1$, and its angular width is given by $\Delta \theta = \theta_2 - \theta_1$. We also use $\theta_p = (\theta_1 + \theta_2)/2$ to indicate the angular position of the centre of the ASM in the circle of radius $(r_1 + r_2)/2$ as shown in Fig. 5.1. The magnetization vector $\mathbf{M}$ could be pointing in either the radial or tangential direction as shown in Fig. 5.21. Furthermore, the radial direction could point toward the centre of the system (i.e., $\mathbf{M} = -|\mathbf{M}|\mathbf{u}_r$ for $A_1$) or outward the centre of it (i.e., $\mathbf{M} = +|\mathbf{M}|\mathbf{u}_r$ for $A_2$). Similarly, the tangential magnetization could be in the clockwise direction (i.e., $\mathbf{M} = -|\mathbf{M}|\mathbf{u}_\theta$ for $A_3$) or in the counterclockwise direction (i.e., $\mathbf{M} = +|\mathbf{M}|\mathbf{u}_\theta$ for $A_4$). $\mathbf{u}_r$ and $\mathbf{u}_\theta$ represent the unit vectors in a cylindrical coordinate system and $|\mathbf{M}|$ represents the magnetization grade of the permanent magnet. Finally, the notation $A_i^{\theta_p}$ is used to indicate that the centre of $A_i$ is located at the angular position given by $\theta_p$. For instance, $A_1^{180^0}$ indicates that the ASM $A_1$ is centred at the angular position of $180^0$. 

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In a real application, the IPM can be off the centre and tilted as it will move along with the CE. However, for the sake of simplicity, the optimization of the driving magnetic system is carried out by assuming that it can only rotate about the Z axis and that the IPM is concentric with the driving magnetic system \[111\]. The IPM’s centre is also located at point P and can freely rotate about the Z axis. The assessment of the magnetic torque imparted to the IPM and how it is affected by changes in the IPM’s location and orientation are later presented in Chapters 6 and 7.

### 5.2.2 Analytical models

Since the magnetic torque \( \tau_z \) imparted to the IPM, which is described by Eq. 3.16, is proportional to the magnitude of \( \mathbf{B} \) (i.e., \( |\mathbf{B}| \)), an improvement in \( |\mathbf{B}| \) will increase the magnitude of \( \tau_z \). Therefore, we aim to maximize \( |\mathbf{B}| \) at the centre of the system that also coincides with the centre of the IPM. To this end, we use analytical models to compute \( |\mathbf{B}| \) and to optimize the dimensions of the ASMs (i.e., \( \Delta r, \Delta \theta, \Delta z \)), since these three variables affect \( \mathbf{B} \). Specifically, we use two analytical models and compare them with experimental results to determine the most accurate three-dimensional model to calculate \( |\mathbf{B}| \) at the centre of the system. Once we select the appropriate analytical model, based on its accuracy, we use it to optimize the dimensions of the driving magnetic system.

The first analytical model, called Model1, is based on the Coulombian model for uniformly magnetized tile permanent magnets \[106\] and the second analytical model, called Model2, is based on the Amperian current model for radially magnetized tile permanent magnets \[85, 117\] and for tangentially magnetized tile permanent magnets \[118\]. Please note that Model1 and Model2 defined in Section 3.6.2 shall not be confused with Model1 and Model2 defined in Section 5.2.2 because the former one refer to models for the magnetic torques whereas the latter ones refer to analytical models for the magnetic flux densities. For the sake of brevity, these analytical models to compute \( \mathbf{B} \) are not presented in this thesis but are available in the literature \[85, 106, 117, 118\]. These 3D analytical models are expressed in cylindrical coordinates as

\[
\mathbf{B}(r, \theta, z) = B_r(r, \theta, z)\mathbf{u}_r + B_\theta(r, \theta, z)\mathbf{u}_\theta + B_z(r, \theta, z)\mathbf{u}_z \quad [\text{T}] \tag{5.1}
\]
Each component is a scalar function of the dimensions of the ASMs. However, we are only interested in the radial component $B_r$ and in the tangential component $B_\theta$, since only these two components will tend to rotate the IPM about the Z axis. We compare these two models to estimate the magnetic flux density at the centre of the system (x=y=z=0). Since we aim to determine the most accurate model between Model1 and Model2, we can choose to compare either $B_r$ or $B_\theta$ at the centre of the system. The next two subsections present these comparisons for $B_r$ produced by radially and tangentially magnetized ASMs when their dimensions are changed. All these results obtained from the analytical models were programmed in Matlab.

### 5.2.2.1 Radially magnetized arc-shaped permanent magnet

We consider in Fig. 5.22 a radially magnetized ASM $\Lambda_{180^0}$ with magnetization grade $|\mathbf{M}|$ of 1.32 T (i.e., N45). We have taken the following dimensions in Fig. 5.22 (a): $\Delta r = r_2-r_1$ with $r_1=30$ mm and $30$ mm$<r_2<330$ mm, $\Delta z=30$ mm with $z_2=15$ mm and $z_1=-15$ mm, $\theta_1=165^0$ and $\theta_2=195^0$. The dimensions used in Fig. 5.22 (b) are $\Delta r=20$ mm with $r_1=30$ mm and $r_2=50$ mm, $z_2=\Delta z/2$ mm and $z_1=-\Delta z/2$ mm with $0<\Delta z<300$ mm, $\theta_1=165^0$ and $\theta_2=195^0$.

![Figure 5.22](image)

Figure 5.22 Comparison of $B_r$ at the centre of the system generated by an ASM $\Lambda_{180^0}$ when (a) only $\Delta r$ varies, (b) only $\Delta z$ varies, (c) only $\Delta \theta$ varies ($B_{\text{max}}=145.12$ mT and it occurs when $\Delta \theta=180^0$ with Model2).
In Fig. 5.22 (c), we show the comparison of $B_r$ when the angular width varies. We have taken the following dimensions: $\Delta r = 20$ mm with $r_1 = 30$ mm and $r_2 = 50$ mm, $\Delta z = 30$ mm with $z_2 = 15$ mm and $z_1 = -15$ mm, $\theta_1 = 180^\circ - \Delta \theta / 2$ and $\theta_2 = \theta_1 + \Delta \theta$ with $0^\circ < \Delta \theta < 360^\circ$.

According to Fig. 5.22 (c), Model2 predicts higher values for $B_r$ than the results from Model1 for $60^\circ < \Delta \theta < 320^\circ$. For $\Delta \theta < 60^\circ$, the results from both models are very similar. These analytical models also predict very similar results for $B_r$ at the centre of the system when changes in $\Delta r$, and $\Delta z$, are made to $A_1^{180^\circ}$ whose angular width is $30^\circ$, as shown in Figs. 5.22 (a) and (b). Since both models predict different results for $A_1^{180^\circ}$ when its angular width $\Delta \theta > 60^\circ$, we want to compare $B_r$ when changes in $\Delta r$, and $\Delta z$, are made to $A_1^{180^\circ}$ for an angular width of $\Delta \theta = 90^\circ$ and for an angular width of $\Delta \theta = 180^\circ$. These results are presented in Fig. 5.23, where we use $r_1 = 30$ mm.

We conclude that these models differ greatly when the angular width is increased. However, one of the advantages of Model1 is that it is a general model that can be used for ASMs with any magnetization direction as long as the angular width is relatively small (approximately $\Delta \theta < 60^\circ$) as it is shown in Fig. 5.22. Furthermore, Figs. 5.22-5.23 suggest that by considering the results of Model2, $B_r$ at the centre of the system is increased if the thickness, length and angular width are increased. However, an increment in such parameters will also increase the volume of the external magnetic system, which should be considered in a realistic application as this external magnetic system is to be moved by motors. Another interesting result from Fig. 5.22 (b) is that an optimal length of about 113 mm maximizes $B_r$ at the centre of the system ($B_{r\text{max}} = 60.3$ mT with Model2). A longer length will not improve $B_r$. 

![Graphs showing comparison of $B_r$ for different models](image1.png) (a)  
![Graphs showing comparison of $B_r$ for different models](image2.png) (b)
Figure 5.23 B_r at the centre of the system created by A_3^{90^\circ} when (a) only Δr varies (Δθ =90^\circ, Δz = 30 mm), (b) only Δr varies (Δθ =180^\circ, Δz = 30 mm), (c) only Δz varies (Δr = 20 mm, Δθ =90^\circ), (d) only Δz varies (Δr = 20 mm, Δθ =180^\circ).

5.2.2.2 Tangentially magnetized arc-shaped permanent magnet

We consider in Fig. 5.24 a tangentially magnetized ASM A_3^{90^\circ} with magnetization grade |M| of 1.32 T (i.e., N45). We have taken the following dimensions in Fig. 5.24 (a): Δr=r_2-r_1 with r_1=30 mm and 30 mm<r_2<330 mm, Δz=30 mm with z_2=15 mm and z_1=-15 mm, θ^Ä =75^\circ and θ^ò =105^\circ. The dimensions used in Fig. 5.24 (b) are Δr=20 mm with r_1=30 mm and r_2=50 mm, z_2= Δz/2 mm and z_1= -Δz/2 mm with 0<Δz <300 mm, θ^Ä =75^\circ and θ^ò =105^\circ.

Figure 5.24 Comparison of B_r at the centre of the system generated by an ASM A_3^{90^\circ} when (a) only Δr varies, (b) only Δz varies, (c) only Δθ varies (B_{max}=78 mT and it occurs when Δθ=180^\circ with Model2).

In Fig. 5.24 (c), we show the comparison of B_r when the angular width varies. We have taken the following dimensions: Δr=20 mm with r_1=30 mm and r_2=50 mm, Δz=30 mm with z_2=15 mm and z_1=-15 mm, θ^Ä =90^\circ-Δθ/2 and θ^ò =θ^Ä +Δθ with 0^\circ<Δθ<360^\circ.

Figures 5.24 (a)-(b) show that both models predict similar results for B_r when changes in the thickness, Δr, and length, Δz, are made to A_3^{90^\circ} whose angular width is 30^\circ.
However, these models differ greatly when the angular width is increased beyond $60^\circ$ as shown in Fig. 5.24 (c). The analytical results of Model2 also indicate that $B_r$ is increased when the dimensions of the ASM are increased. Nevertheless, an angular width larger than $180^\circ$ will not improve $B_r$. This optimal angular width is also obtained for a radially magnetized segment (see Fig. 5.22 (c)).

We also want to compare $B_r$ when changes in $\Delta r$, and $\Delta z$, are made to $A_r^{90^\circ}$ for an angular width of $\Delta \theta = 90^\circ$ and for an angular width of $\Delta \theta = 180^\circ$. These results are presented in Fig. 5.25, where we use $r_1 = 30$ mm. Fig. 5.25 shows that these analytical models differ when the angular width is larger than $60^\circ$.

![Graphs showing comparison of $B_r$ at the centre of the system created by $A_r^{90^\circ}$ under different conditions.](image)

Figure 5.25 $B_r$ at the centre of the system created by $A_r^{90^\circ}$ when (a) only $\Delta r$ varies ($\Delta \theta = 90^\circ$, $\Delta z = 30$ mm), (b) only $\Delta r$ varies ($\Delta \theta = 180^\circ$, $\Delta z = 30$ mm), (c) only $\Delta z$ varies ($\Delta r = 20$ mm, $\Delta \theta = 90^\circ$), (d) only $\Delta z$ varies ($\Delta r = 20$ mm, $\Delta \theta = 180^\circ$).

Since both analytical models differ greatly for radially and tangentially magnetized permanent magnets when their angular widths are larger than $60^\circ$, we compare their theoretical results with experimental results in Section 5.2.3.1 to determine the most accurate model that we can later use to conduct parametric studies and also find the optimal dimensions of the magnetic system that maximizes $|B|$ at the centre of the system.
5.2.3 Optimization of the driving magnetic system

In order to optimize the driving magnetic system by using analytical models, we have set up experiments (as shown in Fig. 5.26) which have allowed us to compare the accuracy of Model1 and Model2. The experimental setup consists of a 3-channel Gauss meter (Lakeshore-Model 460) that was used to measure the magnetic flux density generated by the ASMs. The probe tip of the Gauss meter was mounted on plastic holders which were fabricated with a 3D printer. The probe tip of the Gauss meter can be moved along the X, Y and Z axes. These displacements are controlled by a micromanipulation system based on an X-Y-Z stage.

Figure 5.26 Experimental setups with aluminum magnet cases (Case1 and Case2) to measure $B_r$ when changes in $\Delta \theta$ and $\Delta z$ are made, (a) Case1, (b) Case2, (c) Case1 mounted on the micromanipulation system, (d) Case2 mounted on the micromanipulation system.

5.2.3.1 Accuracy of analytical models

In our experiments with both ASMs $A^{180^\circ}_1$ and $A^{90^\circ}_3$, we used the following dimensions: $\Delta r = 20$ mm with $r_1 = 30$ mm and $r_2 = 50$ mm, $\Delta \theta = 30^\circ$, and $\Delta z = 30$ mm with $z_1 = -15$ mm and $z_2 = 15$ mm. The magnetization grade of each ASM $|\mathbf{M}|$ was 1.32 [T] (i.e., N45). Although any dimensions and magnetization grade can be chosen to verify the accuracy of the two analytical models, we use these specific dimensions and magnetization grade because they are commercially available ASMs. With these dimensions, different arrays are possible by stacking up the segments along the Z axis.
(i.e., increasing $\Delta z$), by placing them one next to the other and thus increasing $\Delta \theta$, or by a combination of increments in both dimensions. For instance, Fig. 5.27 shows the results for $B_r$ generated by arrays of radially magnetized ASMs, while Fig. 5.28 shows the results for $B_r$ created by arrays of tangentially magnetized ASMs.

![Graphs showing $B_r$ results for different conditions](image)

Figure 5.27 $B_r$ at the centre of the system created by $A_{1}^{180^\circ}$ when (a) only $\Delta z$ varies ($\Delta \theta =30^\circ$), (b) only $\Delta \theta$ varies ($\Delta z=30$ mm), (c) only $\Delta z$ varies ($\Delta \theta =90^\circ$).

![Graphs showing $B_r$ results for different conditions](image)

Figure 5.28 $B_r$ at the centre of the system created by $A_{3}^{90^\circ}$ when (a) only $\Delta z$ varies ($\Delta \theta =30^\circ$), (b) only $\Delta \theta$ varies ($\Delta z=30$ mm), (c) only $\Delta z$ varies ($\Delta \theta =90^\circ$), (d) only $\Delta z$ varies ($\Delta \theta =180^\circ$).

The results shown in Figs. 5.27-5.28 indicate that Model2 is more accurate than Model1 in estimating $B_r$ at the centre of the system, although when the angular width
of the ASMs is $30^\circ$, both models predict very similar results as shown in Fig. 5.27 (a) and Fig. 5.28 (a). Based on the accuracy of these analytical models, we use Model2 to conduct parametric studies and the optimization of the driving magnetic system as it is presented in subsections 5.2.3.2-5.2.3.3, respectively.

### 5.2.3.2 Parametric studies

We use Model2 to carry out parametric studies of radially and tangentially magnetized ASMs. The magnetization grade of each ASM $|\mathbf{M}|$ is 1.32 [T] and $r_1=30$ mm. Specifically, we are interested in determining the effects on $B_r$ at the centre of the system due to changes in the three dimensions of the ASMs (i.e., $\Delta r$, $\Delta \theta$, and $\Delta z$).

For an ASM $A_1^{180^\circ}$, we compute $B_r$ at each point of the volumetric region defined by

$30$ mm $\leq r_2 \leq 200$ mm (increments of 10 mm)

$10^\circ \leq \Delta \theta \leq 360^\circ$ (increments of $10^\circ$)

$20$ mm $\leq \Delta z \leq 400$ mm (increments of 10 mm)

The results of $B_r$ are shown in Fig. 5.29. It can be seen that $B_r$ increases with increments of $\Delta r$ and $\Delta z$. $B_r$ also increases with increments of $\Delta \theta$ until $\Delta \theta=180^\circ$. For larger angular widths, $B_r$ will decrease until it reaches zero T.
Figure 5.29 $B_r$ at the centre of the system created by $A_1^{180^\circ}$ when its dimensions are changed: (a) $r_2=40$ mm, (b) $r_2=120$ mm, c) $r_2=200$ mm.

Although, in general, $B_r$ is improved as the dimensions of the ASM are increased (for $\Delta \theta < 180^\circ$), we want to determine the order of priority in which these dimensions should be increased to maximize $B_r$ at the centre of the system. For this purpose, we use an Analysis of Variance (factorial ANOVA) to statistically determine the impact of each dimension on $B_r$ [119]. The full ANOVA results are obtained with Minitab 17 in this study.

For the ANOVA of an ASM $A_1^{180^\circ}$, we use the following region of interest for its three dimensions:

- $35 \text{ mm} \leq r_2 \leq 135 \text{ mm}$ (increments of 5 mm)
- $9.549^\circ \leq \Delta \theta \leq 180^\circ$ (increments of approximately $5^\circ$)
- $5 \text{ mm} \leq \Delta z \leq 105 \text{ mm}$ (increments of 5 mm)

The ANOVA results for these three parameters, which are presented under the column named “Source”, are shown in Table 5.2, where the F value represents the mean square error to residual and is used to determine the significance of each parameter. The P value represents the significance level. Since the ANOVA study is conducted at 5% significance level, when the P value is less than 0.05, the effect of the respective parameter is significant to the response variable which in this case is $B_r$.

<table>
<thead>
<tr>
<th>Source</th>
<th>DF</th>
<th>Adj SS</th>
<th>Adj MS</th>
<th>F-Value</th>
<th>P-Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\Delta r$</td>
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<td>38.39</td>
<td>1.91949</td>
<td>616.2</td>
<td>0.00</td>
</tr>
<tr>
<td>$r_1*\Delta \theta$</td>
<td>18</td>
<td>62.67</td>
<td>3.48193</td>
<td>1117.79</td>
<td>0.00</td>
</tr>
<tr>
<td>$\Delta z$</td>
<td>20</td>
<td>47.66</td>
<td>2.38316</td>
<td>765.05</td>
<td>0.00</td>
</tr>
</tbody>
</table>

As shown in Table 5.2, all parameters have the P-value of less than 0.05. Therefore, the three parameters significantly affect $B_r$ at the 95% confidence interval.
Furthermore, the highest F-value is on the angular width $\Delta \theta$, followed by the F-value on $\Delta z$, and lastly the F-value on $\Delta r$. These F values indicate that to increase $B_r$, it is more effective to firstly increase $\Delta \theta$, followed by increments in $\Delta z$ and the last parameter to be increased is the thickness $\Delta r$ of the ASM.

For an ASM $A_{3}^{90^0}$, we also compute $B_r$ when the three dimensions are changed within the same volumetric region defined for the ASM $A_{1}^{180^0}$, and we also use $r_1=30$ mm. These results are shown in Fig. 5.30.

![Figure 5.30](image)

Figure 5.30 $B_r$ at the centre of the system created by $A_{3}^{90^0}$ when its dimensions are changed: (a) $r_2=40$ mm, (b) $r_2=120$ mm, (c) $r_2=200$ mm.

It can be seen from Fig. 5.30 that $B_r$ increases with increments of $\Delta r$ and $\Delta z$. $B_r$ also increases with increments of $\Delta \theta$ until $\Delta \theta=180^0$. For larger angular widths, $B_r$ will decrease until it reaches zero. Furthermore, the order of priority in which these dimensions should be increased to maximize $B_r$ at the centre of the system is obtained from the analysis of variance. For this analysis of variance, we also use the same region of interest defined for the ANOVA of an ASM $A_{1}^{180^0}$ and the results are presented in Table 5.3.
These $F$ values reported in Table 5.3 indicate that to increase $B_r$, it is more effective to firstly increase $\Delta r$, followed by increments in $\Delta z$ and the last parameter to be increased is the thickness $\Delta \theta$ of the ASM.

<table>
<thead>
<tr>
<th>Source</th>
<th>DF</th>
<th>SS</th>
<th>Adj MS</th>
<th>F-Value</th>
<th>P-Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\Delta r$</td>
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<td>0.62092</td>
<td>485.2</td>
<td>0.00</td>
</tr>
<tr>
<td>$r_2^*\Delta \theta$</td>
<td>18</td>
<td>22.79</td>
<td>1.26588</td>
<td>989.17</td>
<td>0.00</td>
</tr>
<tr>
<td>$\Delta z$</td>
<td>20</td>
<td>23.56</td>
<td>1.17822</td>
<td>920.68</td>
<td>0.00</td>
</tr>
</tbody>
</table>

With these parametric studies and ANOVA results carried out for ASMs $A_1^{180^\circ}$ and $A_3^{90^\circ}$, we determine the effects on $B_r$ at the centre of the system due to changes in their three dimensions and also the order of priority in which they should vary. In the next section (Section 5.2.3.3), we conduct optimization processes to find specific set of dimensions that maximize $B_r$.

### 5.2.3.3 Optimization of the arc-shaped permanent magnets

In this section, we present two optimization processes: the first one aims to maximize $B_r$ for a given constant volume of the ASM ($V_{asm}$), while the second optimization process aims to minimize $V_{asm}$ for a given constraint of desired $B_r$. Since Model2 represents an accurate analytical model that can be used for radially and tangentially magnetized ASMs with arbitrary dimensions and magnetization grade, we present the first optimization process considering the volume $V_{asm}$ of $1.26\times10^{-5}$ m$^3$ as the given constraint. This is a typical volume of a commercial ASM $A_1^{180^\circ}$ (i.e., $r_1=30$ mm, $r_2=50$ mm, $\Delta \theta=30^\circ$ with $\theta_1=165^\circ$ and $\theta_2=195^\circ$, $\Delta z=30$ mm with $z_1=-15$ mm and $z_2=15$ mm, and $|M|=1.32$ [T]). Therefore, we aim to maximize $B_r$ at the centre of the system created by $A_1^{180^\circ}$.

**First optimization process:**
Maximize $f(x)=B_r$
Subject to $h(x)=V_{asm}=1.26\times10^{-5}$ m$^3$

where
\[ f(x): \mathbb{R}^3 \rightarrow \mathbb{R} \]

\[ x = [r_2, \Delta \theta, \Delta z] \]

\( r_1 \) is fixed at 30 mm, but the other dimensions can take an arbitrary value. \( r_2 \) and \( \Delta z \) units are given in mm and the units for the angular width \( \Delta \theta \) are given in degrees. We carry out the following step-by-step procedure:

1. Obtain the isosurface of a constant volume (Fig. 5.31 (a)).
2. Compute \( B_r \) at each point \( x \) (or vertex) that belongs to the isosurface (Fig. 5.31 (b)).
3. Calculate the maximum value of \( B_r \) (i.e., \( B_{r_{\text{max}}} \)) and find \( x_{\text{optimal}} = [r_{2_{\text{opt}}}, \Delta \theta_{\text{opt}}, \Delta z_{\text{opt}}] \) where the maximum occurs.

By following the above procedure, we find: \( B_{r_{\text{max}}} = 51.9 \) [mT], and \( x_{\text{optimal}} = [37.9, 100^0, 27] \). Therefore, \( r_{2_{\text{opt}}} = 37.9 \) mm, \( \Delta \theta_{\text{opt}} = 100^0 \), \( r_1 \ast \Delta \theta_{\text{opt}} = 52.4 \) mm, and \( \Delta z_{\text{opt}} = 27 \) mm. These optimal dimensions for an ASM indicate that the volume of a single commercial ASM of volume \( 1.26 \times 10^{-5} \) m\(^3\) is better distributed by allocating, firstly, more volume to the angular width dimension, followed by volume allocation to the ASM’s length and finally to its thickness, because 52.4 mm > 27 mm > 7.9 mm. These results are in agreement with the ANOVA results in Table 5.2. Furthermore, a single commercial ASM \( A_1^{180^0} \) generates only 37.5 mT at the centre of the system (see Fig. 5.27 (a)), but through this first optimization process we find that the same volume can be optimally distributed to generate a global optimal value of 51.9 mT (an
improvement of about 38%). The inverse optimization process can be carried out to validate if \( B_{\text{max}} = 51.9 \text{ mT} \) is the global maximum.

**Second optimization process:**

Aiming to create \( B_r = 51.9 \text{ mT} \) with an ASM \( A_{180^\circ} \), we attempt to find the minimum volume \( V_{\text{min}} \) (global minimum) required to generate such magnitude of flux density at the centre of the system. If \( V_{\text{min}} = 1.26 \times 10^{-5} \text{ m}^3 \), then we are corroborating again that 51.9 mT is a global maximum (or global optimal).

Minimize \( f(x) = V_{\text{asm}} \)

Subject to \( h(x) = B_r = 51.9 \text{ mT} \)

where

\[
f(x): \mathbb{R}^3 \rightarrow \mathbb{R}
\]

\[
x = [r_2, \Delta \theta, \Delta z]
\]

We carry out the following step-by-step procedure:

1. Obtain the isosurface of a constant magnetic flux density (Fig. 5.32 (a)).
2. Verify if each point \( x \) on the isosurface generates 51.9 mT at the centre of the system. To do this, we compute \( B_r \) at each vertex on the isosurface and obtain Fig. 5.32 (b).
3. Compute \( V_{\text{asm}} \) at each point \( x \) on the isosurface (see Fig. 5.32 (c)).
4. Calculate the minimum value of \( V_{\text{asm}} \) (i.e., \( V_{\text{min}} \)) and find \( x_{\text{optimal}} = [r_{2\text{opt}}, \Delta \theta_{\text{opt}}, \Delta z_{\text{opt}}] \) where the minimum occurs.
Figure 5.32 (a) Isosurface of $B_r=51.9$ mT, (b) $B_r$ generated by each vertex, (c) volume of each vertex and the global minimum volume.

By following the above procedure, we find: $V_{\text{min}}=1.255 \times 10^{-5}$ m$^3$, and $x_{\text{optimal}}=[38, 99^0, 27]$. Therefore, $r_{2\text{opt}}=38$ mm, $\Delta \theta_{\text{opt}}=99^0$, and $\Delta z_{\text{opt}}=27$ mm. These results again confirm the optimal dimensions found from the first optimization process.

We also use the second optimization process to find the optimal dimensions and the minimum volume required to generate a flux density of 37.5 mT with an ASM $A_{180^0}$. We find that the minimum volume $V_{\text{min}}$ of $8.1704 \times 10^{-6}$ m$^3$ can generate 37.5 mT at the centre of the system. This represents an improvement in the volume of the ASM of about 35%. This global minimum volume is reached for a unique set of dimensions: $\Delta z=25$ mm, $\Delta \theta=94^0$, $r_2=36$ mm (and $r_1$ is fixed at 30 mm). By minimizing the volume, we will be able to more easily maneuver the external magnetic system while generating an adequate magnetic field to actuate the slider-crank mechanism embedded in the capsule robot. This is of particular interest when the dimensions of the driving magnetic system are scaled up to actuate the drug release module from an operating distance larger than $r_1=30$ mm which is the operating distance used in our prototype.

In the final parametric study, we progressively increase the volume of the ASM $A_{180^0}$ by multiplying the original volume of a single commercial ASM by a scaling factor. For each volume, we calculate $B_{r\text{max}}$ and $x_{\text{optimal}}$ by following the procedure explained for the first optimization process. Table 5.4 shows the results of a parametric study where the following increments are used: $r_2$: 0.5 mm, increments of $\Delta \theta$: $1^0$, and increments of $\Delta z$: 0.5 mm. By calculating $\Delta r_{\text{opt}}=r_{2\text{opt}}-r_1$ with $r_1=30$ mm, we obtain the results shown in Table 5.4.
From Table 5.4, we can see that the optimal distribution of the volume to generate a maximum $B_r$ is obtained when the volume is allocated firstly along the angular width, secondly along the length and thirdly along the thickness. These results for optimal volume allocation are consistent with the ANOVA results in Table 5.2.

Table 5.4 Variation of $B_{\text{max}}$ due to changes in the volume of the ASM $A_1^{180\degree}$.

<table>
<thead>
<tr>
<th>Scaling Factor [mm]</th>
<th>$\Delta r_{\text{opt}}$ [mm]</th>
<th>$\Delta \theta_{\text{opt}}$ [deg]</th>
<th>$r_1$ [mm]</th>
<th>$\Delta z_{\text{opt}}$ [mm]</th>
<th>$B_{\text{max}}$ [mT]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>7.9</td>
<td>100.0</td>
<td>52.4</td>
<td>27.0</td>
<td>51.9</td>
</tr>
<tr>
<td>2</td>
<td>11.1</td>
<td>112.0</td>
<td>58.6</td>
<td>32.5</td>
<td>84.1</td>
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<td>61.8</td>
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</tr>
<tr>
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<td>38.8</td>
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</tr>
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<td>128.1</td>
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<td>21.0</td>
<td>131.0</td>
<td>68.6</td>
<td>45.2</td>
<td>176.9</td>
</tr>
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<td>8</td>
<td>22.5</td>
<td>133.0</td>
<td>69.6</td>
<td>46.7</td>
<td>189.6</td>
</tr>
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</table>

We also use the first optimization process for an ASM $A_{3}^{90\degree}$ with a commercial volume $V_{\text{asm}}$ of $1.26 \times 10^{-5}$ m$^3$ and $r_1=30$ mm. We find: $B_{\text{max}}=28.5$ mT, and $x_{\text{optimal}}=[36.47, 93^0, 36]$. Therefore, $r_{2\text{opt}}=36.47$ mm, $\Delta \theta_{\text{opt}}=93^0$, and $\Delta z_{\text{opt}}=36$ mm. This $B_{\text{max}}$ is a global maximum. A single commercial ASM $A_{3}^{90\degree}$ generates only 20 mT at the centre of the system (see Fig. 5.28 (a)), but we find that the same volume can optimally be distributed to generate a global optimal value of 28.5 mT (an improvement of about 42.5%).

We also progressively increase the volume of the ASM $A_{3}^{90\degree}$ by multiplying the original volume of a single commercial ASM by a scaling factor. For each volume, we calculate $B_{\text{max}}$ and $x_{\text{optimal}}$ by following the procedure explained for the first optimization process. Table 5.5 shows the results of a parametric study where the following increments are
used: \( r_2: 0.5 \) mm, increments of \( \Delta \theta: 1^\circ \), and increments of \( \Delta z: 0.5 \) mm. By calculating \( \Delta r_{\text{opt}} = r_2 - r_1 \) with \( r_1 = 30 \) mm, we obtain the results shown in Table 5.5.

<table>
<thead>
<tr>
<th>Scaling Factor [mm]</th>
<th>( \Delta r_{\text{opt}} ) [mm]</th>
<th>( \Delta \theta_{\text{opt}} ) [deg]</th>
<th>( \Delta z_{\text{opt}} ) [mm]</th>
<th>( B_{\text{rmax}} ) [mT]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>6.5</td>
<td>93.0</td>
<td>48.7</td>
<td>36.0</td>
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<td>5</td>
<td>14.2</td>
<td>119.0</td>
<td>62.3</td>
<td>57.5</td>
</tr>
<tr>
<td>6</td>
<td>15.5</td>
<td>122.0</td>
<td>63.9</td>
<td>60.5</td>
</tr>
<tr>
<td>7</td>
<td>16.7</td>
<td>123.9</td>
<td>64.9</td>
<td>63.5</td>
</tr>
<tr>
<td>8</td>
<td>17.8</td>
<td>126.1</td>
<td>66.0</td>
<td>66.0</td>
</tr>
</tbody>
</table>

From Table 5.5, we can see that the optimal distribution of the volume is obtained to generate a maximum \( B_r \) when the volume is allocated firstly along the angular width, secondly along the length and thirdly along the thickness. These are the same results obtained from the analysis of variance in Table 5.3.

The results reported in Tables 5.4-5.5 show that, for the same volume, radially magnetized segments always produce higher magnetic flux densities at the centre of the system than the magnetic flux densities produced by tangentially magnetized segments. However, in subsection 5.2.4, we work with the worst scenario, and therefore we only use tangentially magnetized ASMs to experimentally verify the efficacy of the proposed optimization method.

5.2.4 Experimental methods

5.2.4.1 Magnetic flux density

We can use the results depicted in Tables 5.4 and 5.5 to fabricate magnetic systems with optimal dimensions that can maximize \( B_r \) at the centre of the system. For instance,
different arrays can be obtained with five commercial ASMs tangentially magnetized. We present two possible configurations in Figs. 5.33 (a)-(b), which generate the theoretical values of 49.2 mT and 75.3 mT at the centre of the system, respectively. However, for the same volume, we obtain from Table 5.5 the optimal dimensions (shown in Fig. 5.33 (c)) of a single ASM that can generate a theoretical flux density of 86.8 mT. The magnetic structure shown in Fig. 5.33 (a) has a poor distribution of its volume along its three dimensions, and that is the reason why it only generates 49.2 mT. On the other hand, the magnetic structure shown in Fig. 5.33 (b) has a better distribution of its volume along its three dimensions by allocating most of the volume to its angular width, followed by volume allocation along the length and the smallest dimension given to its thickness. For this reason, this magnetic structure produces a higher flux density of 75.3 mT.

The closer are the dimensions of the magnetic structure to the optimal dimensions, the higher will be the magnetic flux density. The specific optimal dimensions shown in Fig. 5.33 (c) could be customized by a manufacturer. However, we have decided to implement the magnetic structure shown in Fig. 5.33 (b) to reduce costs and used the symmetry of the system to double the magnitude of the flux density at the centre of the system. This implementation is shown in Fig. 5.34 (a) with off-the-shelf ASMs.

If we customized the magnetic structure with two tangentially magnetized segments $A_{3}^{90^\circ}$ and $A_{4}^{270^\circ}$, each segment with the optimal dimensions presented in Table 5.5
(using the scaling factor 5) and also depicted in Fig. 5.33 (c), we would obtain the flux density $B_x$ along the X axis as shown in Fig. 5.34 (b) (the black line), where $B_x=86.8*2=173.6$ mT at the centre of the system (note that $B_x=B_r$ along the X axis).

![Image](image1.png)

Figure 5.34 (a) Magnetic structure made of only tangentially magnetized segments of the types $A_3$ and $A_4$, (b) $B_x$ along the X axis created by: an optimal magnetic system (black line) and the practical magnetic system shown in Fig. 5.34 (a) (theoretical results using Model2: green line, and experimental results: the dotted red line), (c) Vector field of the magnetic flux density norm on the plane $z=0$ generated by the structure shown in Fig. 5.34 (a).

For practical reasons, we assembled the magnetic structure shown in Fig. 5.34 (a) and we measured $B_x$ along the X axis as shown in Fig. 5.34 (b) (the dotted red line), where $B_x=144.2$ mT at the centre of the system. The optimal dimensions of the magnetic system would generate an approximately constant $B_x$ in the range $-17$ mm $< x < 17$ mm which is advantageous to guarantee a stable transmitted peak torque on the IPM regardless of its position within that range of operation. On the other hand, the practical assembly of the magnetic system generates a U-shape $B_x$ curve with a minimum experimental value of 144.2 mT at $x=0$. This result from the practical assembly indicates that the IPM will experience a minimum peak torque if it is located...
at the centre of the system, but if it’s moved from the centre, the transmitted peak torque will increase proportionally.

The analytical results show that $B_y$ is 0 mT along the X axis for the magnetic structure shown in Fig. 5.34 (a). For this reason, we do not show $B_y$ in any results. However, we present in Fig. 5.34 (c) a 2D vector field representation of $|B|$ created by such a magnetic structure in plane $z=0$ (using Comsol). The magnitude of this vector representation was normalized just to show the direction of $B$ around the point P. Fig. 5.34 (c) shows that $|B|$ approximates $B_x$ over a relatively large region around point P (i.e., the centre of the system). Therefore, the $B_x$ component is mainly responsible for the transmitted torque on the IPM in Eq. 3.16. We present in Section 5.2.4.2, the experimental results for the transmitted torque on the IPM with the practical assembly shown in Fig. 5.34 (a).

5.2.4.2 Magnetic torque

A torque gauge (HTG2-40 supplied by IMADA) with its respective torque sensor held in place a 3.1 mm cubic IPM with magnetization grade N50 (i.e., $|m|=1.4$ T). The driving magnetic system was mounted on a plastic holder that possesses $30^\circ$ angle indicators and allows its manual rotation about the Z axis. This driving magnetic system, once mounted on the plastic holder, could be moved along the Y and Z axes. These displacements were controlled by a micromanipulation system based on an X-Y-Z stage as shown in Fig. 5.35 (a).

In the first experiment, we positioned the IPM's centre at the centre of the system with its magnetization vector $m$ aligned with the X axis (see Fig. 5.35 (a)), and manually rotated the driving magnetic system about the Z axis every $30^\circ$ until a full cycle was completed. The theoretical and experimental results of the transmitted torque on the IPM $\tau_x$ are shown in Fig. 5.35 (b). It can be seen that the peak torque is transmitted to the IPM when the misalignment angle $\gamma$ between $m$ and $B$ reaches $90^\circ$. The theoretical results were estimated with Eq. 3.16 where $B$ is calculated using Model2. In the second experiment, the IPM's centre was moved along the X axis with increments of 3 mm and its magnetization vector $m$ was aligned with the Y axis along the entire trajectory. In this second experiment, the driving magnetic system was never rotated to
guarantee that $\gamma = 90^0$ and a peak torque were transmitted at all times. The results from the second experiment are shown in Fig. 5.35 (c).

![Image of experimental setup](image)

Figure 5.35 (a) Experimental setup to measure the transmitted torque to the cubic IPM by the driving magnetic system made of 5 ASMs $A_3$ and 5 ASMs $A_4$, (b) $\tau_z$ imparted to the IPM (an experimental peak torque of 5 mNm is reached at $\gamma = 90^0$), (c) Peak torque transmitted to the IPM as its centre is moved along the X axis.

These experimental results for the transmitted torque on the IPM show that the minimum peak torque of 5 mNm is obtained when the IPM’s centre coincides with the centre of the system, and the peak torque is further improved if the IPM’s centre is located at any other position in the X axis. If the driving magnetic system were customized with two tangentially magnetized ASMs, each with the optimal dimensions shown in Fig. 5.33 (c), we would obtain an approximately constant theoretical peak torque of 6 mNm in the range $-17 \text{ mm} < x < 17 \text{ mm}$ because the IPM would be under an approximately uniform $B_x$ of 173.6 mT in the same range. If the IPM’s centre was located outside that range of operation (i.e., $x < -17 \text{ mm}$ or $x > 17 \text{ mm}$), the peak torque
would decrease and the driving magnetic system may need to be repositioned so that the IPM’s centre can fall again within the adequate region of operation.

Although these driving magnetic systems made of only tangentially magnetized segments can be fabricated by assembling commercially available ASMs or by customizing the ASMs with optimal dimensions, in either case, the imparted peak torque to the IPM is at least 5 mNm within the region of operation. However, two additional facts should be considered when this magnetic system is scaled up [111]: i) 5 mNm is more than enough peak torque to actuate the piston of the drug release module, knowing that 3.5 mNm is sufficient for the release, ii) a peak torque is not always required to actuate the piston. For instance, magnetic torques of 2 and 4 mNm, which can be obtained when γ=30° and γ=60°, respectively, are also adequate to release a variety of drug compounds. Furthermore, if the driving magnetic system was fabricated by only assembling radially magnetized ASMs, the peak torque on the IPM at the centre of the system will be higher than the peak torque generated with the driving magnetic system made of only tangentially magnetized ASMs because radially magnetized ASMs can produce higher flux densities as suggested by the results presented in Tables 5.4-5.5.

We estimate, for example, that with the same volume of 5 commercial segments of the type A₁ and 5 of the type A₂, it is possible to generate $B_x=294.8$ mT at the centre of the system (see Table 5.4 with scaling factor of 5: $B_{r\text{max}}=147.4$ mT, and therefore $B_x=2\times B_{r\text{max}}$). This higher magnetic flux density would allow an increase in the operating distance. Further, this optimal magnetic structure made of only radially magnetized ASMs would impart a peak torque of approximately 10 mNm to the IPM. This higher peak torque would allow a further miniaturization of the IPM. However, if we wanted to impart a peak torque between 5 to 6 mNm with an optimal driving system made of only radially magnetized ASMs, we would select only two segments (A₁ and A₂), each with the optimal dimensions presented in Table 5.4 (using the scaling factor 2: $B_{r\text{max}}=84.1$ mT, and therefore $B_x=168.2$ mT=$2\times B_{r\text{max}}$). This selection implies a reduction of 60% ($\frac{5-2\times100}{5}$%) in the volume if compared with the optimal dimensions of 2 tangentially magnetized ASMs, each with the dimensions shown in Fig. 5.33 (c). These results clearly indicate that the driving magnetic system can be scaled up with
optimal dimensions to minimize its total volume while generating adequate magnetic flux densities and magnetic torques to actuate the drug release module embedded in the capsule robot. Consequently, a minimum volume of the driving magnetic system will improve its maneuverability and reduce fabrication costs.

5.2.5 Results and discussion

The optimization of both the IPM and the driving magnetic system is important to obtain an efficient magnetic linkage (i.e., an optimized magnetic field and torque imparted to the IPM) while minimizing the dimensions of the IPM to be embedded in the capsule robot and at the same time minimizing the volume of the driving magnetic system to improve its maneuverability and reduce its fabrication cost. Furthermore, the size optimization of the driving magnetic system not only helps to minimize its volume, but also allows larger operating distances to actuate the IPM and enables further miniaturization of the IPM.

In this section the size optimization of the driving magnetic system which consists of an array of ASMs is presented. Specifically, we have found optimal dimensions for the driving magnetic system (i.e., thickness, angular width and length) and obtained an optimized magnetic field and subsequently, a magnetic torque. This was carried out by using a very accurate analytical model, called Model2, which allows a fast global optimization and is useful for any arbitrary dimension of the ASM. Due to its high accuracy, Model2 can be used to scale up the driving magnetic system which is necessary for the final application where larger operating distances are needed. We have found that Model1 was not accurate in predicting the flux density if the angular width of the ASM was larger than 60°.

We have also found, through parametric studies and a statistical analysis (ANOVA), efficient ways to distribute the volume of the ASMs. Specifically, we have found that for both radially and tangentially ASMs, it is always more efficient to firstly increase $\Delta \theta$, followed by increments in $\Delta z$ and the last parameter to be increased is the thickness $\Delta r$. In this order of priority, the volume can be minimized while obtaining higher flux densities and magnetic torques in the centre of the system where the IPM is located. Our results also indicate that optimal radially magnetized ASMs always
generate higher flux densities than what can be generated with optimal tangentially magnetized ASMs. Although, in Section 5.2, we have presented driving magnetic systems made of segments with only one type of magnetization direction (either radially or tangentially magnetized ASMs), it is also possible to fabricate driving magnetic systems with a combination of both types of ASMs as it is presented in the next section.

5.3 Size optimization with combination of ASMs

In this section, we conduct the size optimization of the external magnetic system by considering a combination of the radially and tangentially magnetized ASMs. This provides remarkable flexibility in enlarging the distance from the external system to the target point (the IPM) or minimizing the volume of the target point, which is crucial for a magnetic device like a robotic capsule operating within the human body. This optimization methodology can be applied to any magnetic propulsion system requiring a remote transmission of a rotational magnetic field or torque to a target device.

5.3.1 External magnetic system

Our aim in this section is to optimize the dimensions of the ASMs (i.e., thickness: $\Delta r$, angular width: $\Delta \theta$, and length: $\Delta z$) to enhance the magnetic field at the centre of the system where the IPM can be located. In Section 4.4.3 we have found that 4 ASMs can generate 113 mT at the centre of the system and impart a magnetic torque of 3.5 mNm to a 3.1 mm cubic IPM. We then optimized the external magnetic system with 12 off-the-shelf ASMs in Section 5.1 to increase the magnetic flux density to 303 mT and the magnetic torque to the same cubic IPM was increased to 10 mNm. In subsection 5.2.4.2, we have found that with only 10 ASMs radially magnetized (with the same volume of 10 off-the-shelf ASMs but with optimal dimensions), it is possible to obtain 294 mT and nearly 10 mNm on the same cubic IPM. Therefore, we aim, in this section, to generate more than 303 mT at the centre of the system by optimizing the dimensions of the external magnetic system made of both radially and tangentially magnetized ASMs as shown in Fig. 5.21 (a).
For the sake of simplicity, the optimization of the external magnetic system is carried out by assuming that it can only rotate about the Z axis. In subsection 5.3.2, we use Model2 which is the analytical solution that is an accurate enough model for the size optimization of ASMs. This analytical model allows the computation of the magnetic flux density $B$, which is expressed in Eq. 5.1, and is based on the Amperian current model for radially magnetized tile permanent magnets [85, 117] and for tangentially magnetized tile permanent magnets [118].

5.3.2 Optimization of the external magnetic system

A combination of off-the-shelf radially and tangentially magnetised ASMs is used in Section 5.1 where we have found the optimal angular positions $\theta_p$ for each of the ASMs [107]. Specifically, we have found that if the centres of $A_1$ and $A_3$ are located in the range $90^0 \leq \theta_p \leq 180^0$ (i.e., the second quadrant), we can generate optimal values of $B_r$ at the centre of the system ($B_\theta = 0$ at point P). Therefore, we aim to maximize $B_r$ by using two ASMs located in the second quadrant as shown in Fig. 5.36. Once the optimal dimensions of the ASMs that maximize $B_r$ are found, we use the symmetry of the system and the superposition principle to obtain the total dimensions of the external magnetic system in the entire range of $360^0$.

![Figure 5.36 Top view of the combination of radially and tangentially magnetized ASMs located in the second quadrant. Note: $B_r= B_x$ at the centre of the system.](image)

The internal radius of the external magnetic system, $r_{in}$, is kept constant to guarantee that the ASMs cannot go closer to the centre of the system (i.e., $r_{11}=r_{31}=r_{in}$), thus a maximum radial operating distance is maintained. We vary the external radii $r_{12}$, $r_{32}$, the angular widths $\Delta \theta_1$, $\Delta \theta_3$ and the lengths $\Delta z_1$ and $\Delta z_3$. $V_{asm}$ is the volume of a commercially available ASM ($r_1=30$ mm, $r_2=50$ mm, $\Delta \theta=30^0$, $\Delta z=30$ mm) and equals $1.2566 \times 10^{-5}$ m$^3$. The optimization problem is described as follows:

Maximize $f(x) = B_r(x_1) + B_r(x_3)$ [T]
Subject to $h(x) = V_{\text{totalasm}} = V_{\text{asm}1}(x_1) + V_{\text{asm}3}(x_3)$ [m$^3$]

And $\Delta \theta_1 + \Delta \theta_3 = 90^\circ$

where

$$f(x): \mathbb{R}^3 \to \mathbb{R}$$

$$x = [r_2, \Delta \theta, \Delta z]$$

and $x_1 = [r_{12}, \Delta \theta_1, \Delta z_1]$ and $x_3 = [r_{32}, \Delta \theta_3, \Delta z_3] \in \mathbb{R}^3$. $r_2$ and $\Delta z$ units are given in mm and the units for the angular width $\Delta \theta$ are given in degrees. $V_{\text{totalasm}}$ is the total volume given as a constant value.

The following steps are used in the optimization procedure:

1. Evaluate $B_{r3}$ at each point $x_3$
2. Calculate $V_{\text{asm}1} = V_{\text{totalasm}} - V_{\text{asm}3}(x_3)$
3. Obtain the isosurface of a constant volume $V_{\text{asm}1}$
4. Compute $B_{r1}(x_1)$ at each point $x_1$ (or vertex) that belongs to the isosurface and such $\Delta \theta_1 + \Delta \theta_3 = 90^\circ$
5. Calculate the maximum value of $B_{r1}$ (i.e., $B_{r1\text{max}}$)
6. Compute $f(x)$ by adding $B_{r3}$ and $B_{r1\text{max}}$ for each point $x_3$
7. Calculate the maximum value of $f(x)$ and the corresponding optimal dimensions $x_{1\text{opt}} = [r_{12\text{opt}}, \Delta \theta_{1\text{opt}}, \Delta z_{1\text{opt}}]$ and $x_{3\text{opt}} = [r_{32\text{opt}}, \Delta \theta_{3\text{opt}}, \Delta z_{3\text{opt}}]$ where the maximum occurs.

### 5.3.2.1 Optimal magnetic systems

We start the first optimization process by selecting an external magnetic system with a total volume of $12^* V_{\text{asm}}$ which represents the total volume of 12 commercially available ASMs that can be distributed in the entire range of $360^\circ$. Therefore, we use $V_{\text{totalasm}} = 3^* V_{\text{asm}}$ in the optimization procedure, which represents one fourth of $12^* V_{\text{asm}}$ that would be allocated to the second quadrant, and $r_n = 30$ mm. The magnetization grade of these commercially available ASMs, which we have also used in previous experimental sections, is 1.32 T (i.e., N45). We obtain the global maximum value of $f(x) = B_{r1\text{max}} = 87.6$ mT with the optimal dimensions of the ASMs $x_{1\text{opt}} = [47.99, 57, 41]$ and $x_{3\text{opt}} = [40, 33, 45]$. These optimal dimensions are shown in Fig. 5.37 a).
Figure 5.37 Optimal dimensions in the range of $90^\circ$ with a total volume of (a) $3^\star V_{asm}$ and (b) $6^\star V_{asm}$.

By using the superposition principle, the combination of the four quadrants, with a total volume of $12^\star V_{asm}$ distributed along the circle of radius $r_m=30$ mm, would give a global maximum value of $B_{r_{max}}=4^\star 87.6=350.4$ mT at the centre of the system, as shown in the optimal curve depicted in Fig. 5.38 (c). Although we have previously obtained a local optimal value of $B_r=318.5$ mT with the same volume of $12^\star V_{asm}$ [107], we can see now that the global maximum value represents an improvement of 10%. This additional magnetic flux density of 31.9 mT generated by the optimized external magnetic system could be used, for example, to actuate additional on-board modules if we consider that a minimum value of 11.2 mT was used to achieve drug delivery in human blood vessels [54].

In our second optimization process, we choose an external magnetic system with a total volume of $24^\star V_{asm}$ which represents the total volume of 24 commercially available ASMs, therefore we use $V_{totalasm}=6^\star V_{asm}$ in the optimization procedure, which represents one fourth of $24^\star V_{asm}$ that would be allocated to the second quadrant. We obtain the global maximum value of $f(x)=B_{r_{max}}=128$ mT for the optimal dimensions of the ASMs $x_{1_{opt}}=[55.43, 57, 50]$ and $x_{3_{opt}}=[46, 33, 61]$. These optimal dimensions are shown in Fig. 5.37 (b). The combination of the four quadrants, with a total volume of $24^\star V_{asm}$ distributed along the circle of radius $r_m=30$ mm, would give a global maximum of $B_{r_{max}}=4^\star 128=512$ mT at the centre of the system as shown in the optimal curve depicted in Fig. 5.38 (d).

5.3.2.2 Practical magnetic systems

In our previous Section 5.2, we have shown that to increase $B_r$, it is more effective to firstly increase $\Delta \theta$, followed by increments in $\Delta z$ and the last parameter to be increased is the thickness $\Delta r$ of the ASMs [120]. Therefore, it may be more practical to firstly fix
the values of $\Delta z$ and $\Delta r$ and only change the angular widths of $A_1$ and $A_3$ simultaneously.

For example, by fixing $r_{in}$=30 mm, $r_{out}$=50 mm and $\Delta z = 30$ mm, we find the following optimal angular widths in the second quadrant: $\Delta \theta_3 = 28^0$ and $\Delta \theta_1 = 62^0$. These dimensions would generate a theoretical value of $B_{r\max} = 82.4$ mT. However, these angular widths must be customized. In practice, we can approximate these angular widths by selecting $\Delta \theta_3 = 30^0$ and $\Delta \theta_1 = 60^0$ since an off-the-shelf ASM has an angular width of 30$^0$ and a multiple of this angular width is possible to assemble. With these practical dimensions, we would obtain $B_r = 82.3$ mT. By considering the combination of the four quadrants, we can assemble a practical magnetic system with off-the-shelf ASMs as shown in Fig. 5.38 (a) that would generate a total flux density of $B_r = 4\times 82.3 = 329.2$ mT at the centre of the system as shown in the practical curve depicted in Fig. 5.38 (c). This value is 21.2 mT less than the global optimal value of 350.4 mT which is generated by the specific dimensions shown in Fig. 5.37 (a).

Once the first ring of 12 practical ASMs is arranged, we proceed to increase its $\Delta z$ since this increment is preferred over increments in the thickness of the ASMs which tend to produce the least contribution to maximize $B_r$ at the centre of the system. To this end, we fix $r_{in}$=30 mm, $r_{out}$=50 mm and increase $\Delta z$ from 30 mm to 60 mm. With these practical dimensions, we find the following optimal angular widths in the second quadrant: $\Delta \theta_3 = 32^0$ and $\Delta \theta_1 = 58^0$; these dimensions would generate a theoretical value of $B_{r\max} = 124.8$ mT. However, these angular widths must be customized by a manufacturer. In practice, we can approximate these angular widths by selecting $\Delta \theta_3 = 30^0$ and $\Delta \theta_1 = 60^0$. With these practical dimensions, we would obtain $B_r = 124.7$ mT.

By considering the combination of the four quadrants, we can assemble a practical magnetic system with off-the-shelf ASMs as shown in Fig. 5.38 (b) that would generate a total flux density of $B_r = 4\times 124.7 = 498.8$ mT at the centre of the system as shown in Fig. 5.38 (d). This value is 13.2 mT less than the global optimal value of 512 mT which is generated by the specific dimensions shown in Fig. 5.37 (b). This practical magnetic system is implemented with 24 off-the-shelf ASMs and shown in Fig. 5.39. This is the final optimal magnetic system that we use in Chapters 6-7 to actuate a small IPM embedded in the prototype of capsule robot.
Figure 5.38 Practical magnetic systems made of arrays of ASMs (\( r_{in} = 30 \text{ mm}, r_{out} = 50 \text{ mm}, \Delta\theta = 30^\circ, \Delta z = 30 \text{ mm} \)), (a) assembly with 12 ASMs, (b) assembly with 24 ASMs; comparison of \( B_x \) generated by optimal and practical magnetic systems each with a total volume of: (c) 12\( *V_{asm} \) (i.e., \( \Delta z = 30 \text{ mm} \) in the practical assembly), (d) 24\( *V_{asm} \) (i.e., \( \Delta z = 60 \text{ mm} \) in the practical assembly). Note: \( B_r = B_x \) along the X axis.

Figure 5.39 Optimal practical assembly of the external magnetic system.
5.4 Conclusions

In this chapter, we have optimized the external magnetic system to enhance the magnetic field and the transmitted torque on a 3.1 mm cubic IPM that has been placed at the centre of the system. We have optimized the angular position and size of each ASM, and have assembled a practical magnetic system with its dimensions close to the optimal dimensions obtained through our proposed optimization methodology. This final optimal external magnetic system was designed to generate approx. 500 mT at the centre of the system, using the minimum possible volume, and has been built with 24 off-the-shelf ASMs.

Our scaled down magnetic system made of off-the-shelf ASMs has been a useful platform to validate our theoretical results and we also use it in Chapters 6 and 7. Since the theoretical results are based on the analytical functions that are valid for any size of the ASMs, then these results from Chapter 5 are valid when we also scale up the external magnetic system. Further details of the scaling laws have been provided in Chapter 6 (more specifically in subsection 6.1.1). However, the specific scaled up external magnetic system needs to be further assessed (for the final implementation in a real medical application) in terms of its practical maneuverability as we expect that the ASMs would be mounted on a mobile platform that will be powered by electric motors. The power demand on the motors and the technical specifications of the mobile platform that would be controlled via a joystick are out of the scope of this thesis. Nevertheless, there is no doubt that an external magnetic system with a minimum weight is highly desirable to ease the maneuverability and decrease the power demand on the motors. Therefore, our optimization results presented in this chapter clearly contribute to these objectives.

Now that the external magnetic system has been optimized and implemented, we focus on the analysis of the magnetic torque on the IPM whose centre can be located at any position and with arbitrary orientation within the cylindrical region of operation. These analyses will be conducted in Chapters 6-7.
Chapter 6
Analysis of the Magnetic Torque on a Tilted IPM Constrained by a Region of Operation

6.1 Changes in the IPM’s location

The IPM can be located at any position within the cylindrical region of interest defined by \( r_{in} = 30 \) mm and the length of \( \Delta z = 60 \) mm (see Fig. 5.38 (b) for the dimensions of the final external magnetic system). Therefore, in this section, we assess the magnetic torque due to changes only in the IPM’s location within the cylindrical region of interest.

The IPM’s axial axis \( Z’ \) remains parallel to the \( Z \) axis of the system but arbitrary orientations of the IPM are presented in Section 6.2 and Chapter 7 (see Section 3.2 for the definition of the coordinate systems). For this purpose, we firstly analyze changes in the IPM’s location in the plane \( z = 0 \) and finally changes in the IPM’s location as its centre moves along the \( Z \) axis. To do this, we investigate how the magnetic flux density \( B \), described by Eq. 5.1, varies in the region of interest (i.e., the plane \( z = 0 \) and for points along the \( Z \) axis). We do this because according to Eq. 3.1, the magnetic torque \( \tau \) exerted on the IPM depends on \( B \).

\( B \) represents a static magnetic flux density that becomes a rotating magnetic flux density when the external magnetic system is rotated about the \( Z \) axis by an angle \( \theta_{EPM} \) (see Fig. 3.1). When the ASMs rotate about the \( Z \) axis, only the radial and tangential components of \( B \) will contribute to the torque \( \tau_{zr} \) imparted to the IPM. The relative misalignment between \( m \) and \( B \) causes the IPM to rotate about its \( Z’ \) axis, tending to align \( m \) with \( B \). Since the rotation of the IPM is caused by the relative misalignment of those two vectors, we can choose to fix the direction of \( m \) and rotate \( B \) (by varying \( \theta_{EPM} \)) or viceversa. We choose the latter to derive \( \tau_{zr} \) as the IPM’s location changes within the 3-dimensional region of interest. Therefore, for the purpose of the analysis of \( \tau_{zr} \), we fix the ASMs (i.e., \( \theta_{EPM} = 0^0 \), creating a static magnetic flux density) and move the IPM’s centre along circular trajectories, which are described by the parameters \(( r, \theta )\). The IPM’s magnetization vector \( m \) is always aligned with \( u_r \) (i.e., \( m = |m| u_r \)). In this case \( \tau_{zr} \), which is the axial component of \( \tau \) described by Eq. 3.1, is expressed as

\[
\tau_{zr}(r, \theta, z) = V |m| B_\theta(r, \theta, z)/\mu_0 \tag{6.1}
\]
This indicates that only the tangential component $B_{\theta}$ will tend to rotate the IPM about its $Z'$ axis when the IPM’s centre is moved along circular trajectories while maintaining $\mathbf{m} = |\mathbf{m}| \mathbf{u}_r$. In other words, only the component of $\mathbf{B}$ that is perpendicular to $\mathbf{m}$ will contribute to $\tau_{zz'}$. According to Eq. 6.1, $\tau_{zz'}$ will have the same trend as $B_{\theta}$. Thus, we analyze $B_{\theta}(r, \theta, z)$ for all the possible positions of the IPM within the region of interest while $\mathbf{m}$ is aligned with $\mathbf{u}_r$. For the first analysis, we estimate $B_{\theta}$ at $z=0$ by varying $r$, and $\theta$. This theoretical result is shown in Fig. 6.1. (a). It can be seen that $B_{\theta}$ reaches its peak value when $\theta = 90^0$, varying from 499 mT at $r=0$ and increasing its value to approx. 607 mT at $r=24$ mm. In other words, $B_{\theta}$ is maximum when the ASMs are rotated by $\theta = 90^0$ (at $z=0$, and $0 \leq r \leq 24$ mm). Please note that to move the IPM’s centre along circular trajectories $(r, \theta)$ with its $\mathbf{m}$ aligned with $\mathbf{u}_r$ at all times while the ASMs are not rotated, (i.e., $\theta_{\text{EPM}}=0^0$) is equivalent, for the purpose of the analysis of $\tau_{zz'}$, to fixing $\mathbf{m}$ with $\mathbf{u}_x$ (i.e., $\theta=0^0$) and rotating the ASMs by varying $\theta_{\text{EPM}}$. Therefore, in this section, $\theta$ and $\theta_{\text{EPM}}$ are interchangable.

In the final analysis, we estimate $B_{\theta}$ at $\theta = 90^0$ by varying $r$ and $z$. This theoretical result is shown in Fig. 6.1 (b). It can be seen that $B_{\theta}$ reaches its peak value when $z=0$, varying from 499 mT at $r=0$ and increasing its value to approx. 607 mT at $r=24$ mm. However, $B_{\theta}$ decreases as we move away from $z=0$. Specifically, $B_{\theta}$ decreases to 291 mT at $r=0$, $z=\pm30$ mm.

![Figure 6.1 B_\theta generated by the practical assembly shown in Fig. 5.38 (b), (a) B_\theta at z=0, (b) peak values of B_\theta for r and z displacements.](image)

These theoretical results indicate that a maximum $B_{\theta}$, and subsequently a maximum $\tau_{zz'}$, can be obtained if the IPM’s centre is located in the plane $z=0$ and moved away
from the centre of the system (i.e., point P). These results also confirm that the point P represents a critical point for the transmitted torque and thus its selection in the optimization processes carried out in Chapters 3-5 is appropriate. These theoretical results are verified in the experimental subsection (subsection 6.1.2).

6.1.1 Scaled-up magnetic system

When the dimensions of the practical magnetic system shown in Fig. 5.38 (b) are scaled up by a factor $s$ and the operating distance is simultaneously increased by the same factor, the magnetic flux density $B$ and the magnetic torque $\tau$ remain the same. For example, in our previous work [111] that has also been presented in Chapter 4, we have used the scaling factor of $s=8$ and a larger operating distance of $r_{in}^* s=240$ mm. We have demonstrated that such a scaled-up magnetic system can generate the same magnetic flux density of approximately 114 mT that is obtained with a scaled-down magnetic system that uses an operating distance of $r_{in}=30$ mm with 4 off-the-shelf ASMs. Consequently, any of these two external magnetic systems can produce a sufficient peak torque of approximately 3.5 mNm on a 3.1 mm cubic IPM (N50). This torque actuates a piston with a peak force of 1.67 N that would allow the release of different drug compounds, including liquid and solid forms. Similarly, if we scale up the dimensions of our magnetic system shown in Fig. 5.38 (b) by a factor of 8 and increase the operating distance to 240 mm, we will obtain the same theoretical results for $B_0$ as the ones depicted in Fig. 6.1. For this reason, we do not show $B_0$ generated by the scaled-up magnetic system. It is also expected that $\tau_{xy}$ will remain invariant to the scalability of the external magnetic system because according to Eq. 6.1, the magnetic torque will have the same trend as $B_0$. This indicates that our external magnetic system is well scalable for medical use. This finding is in agreement with similar findings reported in the literature [70, 111] on the scalability of a magnetic actuator while increasing the operating distance with no changes in the magnetic torque on an IPM.

6.1.2 Experimental methods

The experimental setup consisting of a 3-channel Gauss meter and a torque with its respective torque sensor (STJ10Z) held the IPM. A 3.1 mm cubic IPM was connected to the torque sensor via a plastic connector is used. Please note this is the same setup
used in the previous chapters. Both the torque sensor and the probe tip of the Gauss meter can be moved along the X and Z axes and the external magnetic system can be moved along the Y axis and can also be manually rotated about its Z axis by an angle $\theta_{EPM}$. The displacements are controlled by a micromanipulation system constructed of XYZ stages as shown in Fig. 6.2.

![Figure 6.2 Experimental setup consisting of measurement instruments and the practical magnetic system shown in Fig. 5.39.](image)

### 6.1.2.1 Magnetic flux density

In our first experiments, we measured $B_\parallel$ at $z=0$, by varying $r$ and $\theta$. For this purpose, we firstly placed the probe tip of the Gauss meter at the centre of the system and then moved it from $x=0$ until $x=20$ mm (increments of 5 mm) and measured $B_y$ along this trajectory as the ASMs were manually rotated by increments of $\Delta \theta_{EPM}=15^0$. It must be noted that $B_y=B_\parallel$ along the X trajectory, and this is the component of $\mathbf{B}$ that is perpendicular to $\mathbf{m}$ that will contribute to $\tau_z$, as suggested by Eq. 6.1, where $\mathbf{m}=|\mathbf{m}| \mathbf{u}_x = |\mathbf{m}| \mathbf{u}_r$ along the X trajectory. These experimental results, which are depicted in Fig. 6.3, show how $B_\parallel$ changed in the plane $z=0$. 


Figure 6.3 Comparison of analytical and experimental results for $B_\theta$ in the plane $z=0$ generated by the external magnetic system, (a) for $r=0$ and $r=20$ mm, (b) only shows the experimental results for $r=0, 5, 10, 15$ and $20$ mm.

The theoretical value of the magnetic flux density $B_\theta$ at $r=0$, $\theta = 90^0$, and $z=0$ is 499 mT. The corresponding experimental result was measured as 476 [mT] as shown in Fig. 6.3 (b) when $r=0$ and $\theta = 90^0$. This difference of -4.6% is likely due to small misalignments and clearances among the 24 off-the-shelf ASMs that were used in the magnetic system. With reference to the results in Fig. 6.3 (b), $B_\theta$ reached its maximum value at $\theta = 90^0$ for any value of $r$. Furthermore, $B_\theta$ was also enhanced along the axis $\theta = 90^0$ as $r$ was increased, reaching 569 mT at $r=20$ mm.

Since a peak value of $B_\theta$ was always obtained when the ASMs were rotated by $\theta = 90^0$, we measured the peak value of $B_\theta$ by moving the probe tip of the Gauss meter along the X and Z axes. For this purpose, we firstly rotated the ASMs by $\theta = 90^0$, placed the probe tip of the Gauss meter at the centre of the system and then moved it
along the X and Z axes (increments of 5 mm) and measured $B_y$ along these trajectories. These results are shown in Fig. 6.4.

![Graph showing $B_0$ in the plane $\theta = 90^\circ$ for different values of $r$.](image)

Figure 6.4 Comparison of analytical and experimental results for $B_0$ in the plane $\theta = 90^\circ$ generated by the external magnetic system, (a) for $r=0$ and $r=20$ mm, (b) only shows the experimental results for $r=0$, 5, 10, 15 and 20 mm.

The maximum difference between the analytical and experimental results of approximately 6% is likely due to small misalignments and clearances among the 24 ASMs that were used in the magnetic system. With reference to the results in Fig. 6.4 (b), $B_0$ reached its maximum value at $z=0$ for any value of $r$. Furthermore, $B_0$ was also enhanced along the axis $Z=0$ as $r$ was increased, reaching 568 mT at $r=20$ mm.

The experimental results for $B_0$ in Fig. 6.3 (b) and Fig. 6.4 (b) indicate that $B_0$ was enhanced when the ASMs were rotated by $\theta = 90^\circ$ for points in the plane $z=0$ and located away from the centre of the system (i.e., as $r$ was increased). Such points are of interest to obtain an enhanced magnetic torque on the IPM.
6.1.2.2 Magnetic torque

In order to measure $\tau_z$, when the IPM was located at any point in the plane $z=0$, we firstly made the IPM’s centre to coincide with the centre of the system and then moved the IPM along the X axis (from $x=0$ until $x=20$ mm with increments of 5 mm) and its magnetization vector $\mathbf{m}$ was always aligned with the X axis. At each position, we measured $\tau_z$ by manually rotating the ASMs with increments of $\Delta \theta = 15^0$. These experimental results are shown in Fig. 6.5.

For $r=0$, a peak torque of $15.15 \text{ mNm}$ was obtained at $\theta = 90^0$ and $-15.95 \text{ [mNm]}$ at $\theta = 270^0$. For $r=20$ mm, a peak torque of $18.65 \text{ mNm}$ was obtained at $\theta = 90^0$ and $-19.15 \text{ mNm}$ at $\theta = 270^0$. This small increment in the torque can be seen in Fig. 6.5 (c).
when $r$ is changed from 0 to 20 mm. These peak values are also predicted with Eq. 6.1 by considering the experimental values of $B_\theta$ shown in Fig. 6.3. Therefore, these results for points in the plane $z=0$ indicate that the peak torque was increased as the IPM’s centre was moved away from the centre of the system (i.e., as $r$ was increased).

Finally, since a peak torque was always obtained at $\theta = 90^0$ (i.e., when the ASMs were rotated by $\theta = 90^0$), we measured the peak torque as the IPM was displaced along the Z axis (increments of 5 mm). For this purpose, we firstly rotated the ASMs by $\theta = 90^0$, and then moved the IPM along the X axis and along the Z axis. The magnetization vector $\mathbf{m}$ was always aligned with the X axis. These results are shown in Fig. 6.6.

![Graph showing peak torque vs. displacement](image1)

(a)

![3D graph showing peak torque vs. r and z](image2)

(b)
Figure 6.6 Comparison of analytical and experimental results for $\tau_z$ in the plane $\theta = 90^\circ$ generated by the external magnetic system, (a) for $r=0$ and $r=20$ mm, (b) only shows the experimental results for $r=0$, 5, 10, 15 and 20 mm and the optimal region of operation (the red line).

Figure 6.6 shows that the minimum peak torque of 8.65 mNm was obtained for $r=0$ and $z=\pm$ 30 mm. On the other hand, a maximum peak torque of 18.65 mNm was obtained for $r=20$ mm and $z=0$. These experimental results indicate that the optimal region of operation, where maximum peak torques were imparted to the IPM, was located in the plane $z=0$ and as the IPM was moved away from the centre of the system (i.e., as $r$ increases). In this region, the peak torque varied from 15.15 to 18.65 mNm when the IPM’s centre was located at $r=0$, $z=0$ and $r=20$ mm, $z=0$, respectively.

This experimentally determined magnetic torque for the case when the IPM was located at any point within the cylindrical region of interest defined by $r_{in}=30$ mm and the length of $\Delta z=60$ mm, can be predicted by analyzing the theoretical results for $B_\theta$ in the same cylindrical region of interest as it is shown in Fig. 6.1. Therefore, these experimental torque results validate the efficacy of the analytical model described by Eq. 6.1 which is valid for an external magnetic system with any size. Consequently, we conclude that the same magnetic flux density, magnetic torque and optimal region of operation will be obtained if the external magnetic system is scaled up by a factor $s$ and the operating distance is simultaneously increased by the same factor.

Furthermore, the results shown in Fig. 6.6 indicate that the drug release module to be embedded in the capsule robot will always be actuated regardless of the IPM’s location, since a minimum experimental peak torque of 8.65 mN was guaranteed. However, if there is a need to increase the peak torques, the position and orientation of the external magnetic system can be readjusted so that the IPM’s location can fall in the optimal region of operation.

6.1.2.3 Performance of the magnetic linkage

We have shown that a peak torque of approximately 3.5 mNm transmitted onto the same 3.1 mm cubic IPM is sufficient to actuate the drug release mechanism, although a peak torque is not always required to actuate the piston [111]. Therefore, the higher peak torques (from 15.15 to 18.65 mNm) obtained with the practical magnetic system
assembled with 24 off-the-shelf ASMs, the dimensions of which are almost equal to the those of an optimized magnetic system, would allow us to further miniaturize the IPM and/or increase the operating distance. For instance, by using Eq. 6.1, we estimate that the volume of the IPM can be reduced to 1.8 mm$^3$. In this case, for the same magnetization grade, we can obtain the peak torques ranging from 3.2 to 3.9 mNm in the optimal region of operation under the same magnetic system used in this section. The miniaturization of the IPM is important not only to improve the compactness of the drug delivery system (creating more space within the capsule robot to include additional on-board modules) but also to reduce the dimensions of the capsule robot which will help minimize the problem of capsule retention caused sometimes by existing capsule endoscopes [121].

Besides allowing a further miniaturization of the IPM and larger operating distances, the size optimization of the external magnetic system results in a volumetric power density (Vpd) that is well above the power density of many commercially available electric motors (VpdCM) which are listed in Table 6.1. For example, the minimum Vpd obtained for our magnetic actuation system is VpdIPM=0.29 N/mm$^2$ (i.e., 8.65 mNm/3.1$^3$ mm$^3$ when the minimum magnetic torque is transmitted to the IPM).

<table>
<thead>
<tr>
<th>Model</th>
<th>$\varnothing$ [mm]</th>
<th>$L$ [mm]</th>
<th>Max$_\tau$ [mNm]</th>
<th>Vpdr</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Namiki-SBL02</td>
<td>2</td>
<td>5</td>
<td>0.660</td>
<td>7</td>
<td>[122]</td>
</tr>
<tr>
<td>Namiki-SBL04</td>
<td>4</td>
<td>8</td>
<td>5.700</td>
<td>5</td>
<td>[123]</td>
</tr>
<tr>
<td>Faulhaber 0615</td>
<td>6</td>
<td>15</td>
<td>0.240</td>
<td>513</td>
<td>[124]</td>
</tr>
<tr>
<td>Faulhaber 0308</td>
<td>3</td>
<td>8.4</td>
<td>0.030</td>
<td>575</td>
<td>[125]</td>
</tr>
</tbody>
</table>
In Table 6.1, $\Phi$: diameter of the motor, $L$: length of the motor, and $V_{pd}$ is the volumetric power density ratio=$V_{pdIPM}/V_{pdCM}$, with $V_{pdCM}$ as the volumetric power density of the commercial motor. The results presented in Table 6.1 indicate that the size optimization of the external magnetic system is effective in transmitting a higher mechanical power to the IPM. Any additional power can be used to actuate additional on-board mechanisms in the capsule robot such as an anchoring mechanism and/or an active locomotion system. Further considerations about the overall efficiency (e.g., generated power/input power used by the motors to move the scaled up EPMs) can be done once the scaled up external magnetic system is implemented. However, the implementation of this larger system is out of the scope of this thesis and therefore, it is left for future work.

6.1.3 Results and discussion

We have found that further miniaturization of the IPM and larger operating distances are still possible to achieve; thanks to the size optimization of the external magnetic system presented in Section 5.3. Furthermore, our external magnetic system scales well for medical use and provides higher volumetric power densities than commercially available electric motors.

We have also evaluated changes in the magnetic torque due to changes in the IPM’s location within the 3-dimensional region of interest and found an optimal region for the actuation of the drug release mechanism. Although the capsule robot could be actuated even if it was located outside this region of optimal operation, the position and orientation of the external magnetic system could be adjusted to make the capsule robot to operate in the optimal region of actuation. The evaluation of any variation in the magnetic torque when the IPM is tilted at any location within the region of interest,
which is important to estimate for a real medical application, is presented in Section 6.2.

6.2 Changes in the IPM’s orientation: restricted region

Magnetic devices that can remotely be guided and actuated inside biological tissues have become highly attractive due to their potential benefits for minimally invasive procedures [113, 128]. For instance, EPMs have been used to actuate various mechanisms embedded in capsule robots for WCE. However, the actuation of these magnetic devices using EPMs presents challenges for real-time control strategies due to the complexity of the generated magnetic field and its interaction with the IPMs [70]. This complexity has been formally analyzed for the interaction of a single permanent magnet that remotely actuates magnetic devices located at relatively large distances [70, 115, 129]. These studies demonstrate that it is possible to develop real-time control strategies to remotely manipulate magnetic devices that have arbitrary position and/or orientation relative to the single permanent magnet. However, the complexity of the magnetic interactions may increase, for the purpose of achieving real-time control, if the external magnetic system is made of multiple EPMs rather than a single permanent magnet.

Since we have fabricated a prototype of an external magnetic system with 24 off-the-shelf ASMs that we have optimized to generate approximately 476 mT at the centre of the system (x=y=z=0), we aim in this section, to analyze the magnetic torque transmitted to an IPM that is arbitrarily oriented and whose centre can move along the Z axis. Although the IPM’s centre is restricted to only movements along the Z axis, the full analysis of the magnetic torque in the 3-dimensional region of operation for an IPM with arbitrary orientation is presented in Chapter 7. Furthermore, according to the results presented in Fig. 6.6 (b), the minimum peak torques are transmitted to the IPM whose centre moves along the Z axis. Thus, the Z axis represents a critical region where the torque decreases and its selection as the restricted region for the IPM displacement is adequate. Understanding the effects of the IPM’s orientation on its transmitted torque is important for the development of real-time control strategies for magnetic devices. In our specific application of WCE, this analysis of the transmitted torque for a tilted IPM is important because it will allow the clinician to make any
adjustments needed in the external magnetic system for an effective real-time control of the drug delivery system to be embedded in the capsule robot.

6.2.1 Overall system

We align the coordinate system $X_aY_aZ_a$ of the external magnetic system shown in Fig. 5.39 with the general coordinate system $XYZ$ defined in Fig. 3.1. Therefore, in this section, the external magnetic system can only be rotated about the $Z$ axis by an angle $\theta_{EPM}$ (see Fig. 3.1). The capsule robot shown in Fig. 6.7 is to operate within the cylindrical region of interest defined by the maximum radial operating distance $r_1$ of 30 mm and the length of $\Delta z=60$ mm (these are the dimensions of the external magnetic system shown in Fig. 5.39).

![Figure 6.7 the capsule robot with its coordinate system $X'Y'Z'$ located at the centre of the IPM.](image)

A magnetic torque $\tau_{z'}$ (shown in Fig. 6.7) will be imparted to the IPM embedded in the capsule robot as its magnetization vector $\mathbf{m}$ interacts with the rotating magnetic field created by the ASMs. The transmitted torque $\tau_{z'}$ is then converted into a piston force $\mathbf{F}$ by means of a slider-crank mechanism that is physically connected to the IPM, the origin of which coincides with the IPM’s axial axis (i.e., $Z'$) as shown in Fig. 6.7. In a real medical application, the coordinate system $X'Y'Z'$ would change its position and orientation with respect to the general coordinate system $XYZ$ as the capsule robot travels within the digestive system. In this section, the smallest cubic IPM (3.1 mm), with the magnetization grade $|\mathbf{m}|=1.4$ T (i.e., N50) is considered to be placed in a prototype of a capsule robot. When the IPM’s centre coincides with the centre of the external magnetic system and the axes of the coordinate system $X'Y'Z'$ are aligned with the axes of $XYZ$, it is possible to approximately transmit a magnetic peak torque $\tau_{z'}$ of 3.5 mNm]by using only 4 ASMs that generate approximately 114 [mT] at the centre of the IPM [111].
In a real application, the IPM can be off the centre and/or tilted as it will move along with the capsule robot. Understanding the effects of the changes in the IPM’s position and orientation within the cylindrical region of interest on $\tau_{zr}$ is important for the development of real-time control strategies for magnetic devices. In this section, we present this and how $\tau_{zr}$ is affected by changes in the IPM’s orientation (i.e., the axis $Z'$ and $m$ can have arbitrary orientations). Additionally, the IPM’s centre is not only located at the centre of the system but can move along the Z axis.

6.2.2 Theoretical methods

We are interested in estimating and deriving an expression for $\tau_{zr}$ acting on the IPM that is tilted by an angle $\theta_z$ as shown in Fig. 6.8 (a). For this purpose, we choose a coordinate system $X'Y'Z'$ as shown in Fig. 6.8 (b). Therefore, the IPM’s magnetization vector $m$ can rotate about the $Z'$ axis in a circular trajectory with an orientation $\theta_{IPM}$. Thus, with reference to Figs. 6.8 (a)-(b), the orientation of the IPM is fully determined by the parameters $(\theta_{IPM}, \theta_z)$. We can express the projections of $m$ in the plane $X'Y'$ as

$$m = |m|(\cos \theta_{IPM} u_{x'} + \sin \theta_{IPM} u_{y'}) \quad (6.2)$$

Our external magnetic system generates a rotating $B$ at any point along the Z axis as shown in Fig. 6.8 (c). Since $B_z=0$ along the Z axis (for our external magnetic system), when the external magnetic system is rotated by an angle $\theta_{EPM}$, the projection of $B$ in any plane that is parallel to the plane $XY$ can be expressed as

$$B = |B|(\cos \theta_{EPM} u_x + \sin \theta_{EPM} u_y) \quad (6.3)$$

However, we can also find the 3 projections of $B$ in the system $X'Y'Z'$ as

$$B = B_{x'}, u_{x'} + B_{y'}, u_{y'} + B_{z'}, u_{z'} \quad (6.4)$$

where

$$B_{x'} = |B| \cos \theta_{EPM} \cos \theta_z \quad (6.5)$$

$$B_{y'} = |B| \sin \theta_{EPM} \quad (6.6)$$

$$B_{z'} = |B| \cos \theta_{EPM} \sin \theta_z \quad (6.7)$$
The magnetic torque $\tau_{zz}$, which is the axial component of $\tau$ (i.e., the magnetic torque with respect to the system $X'Y'Z'$), is only affected by $B_{x'y'}$ which represents the projections of $B$ in the plane $X'Y'$ and is expressed as

$$B_{x'y'} = B_{x'} u_{x'} + B_{y'} u_{y'} \quad (6.8)$$

By substituting Eq. 6.2 and Eq. 6.8 into Eq. 3.1 and only taking the axial component of $\tau$, we obtain

$$\tau_{z'} = \frac{V}{\mu_0} |m| \left( B_{y'} \cos \theta_{\text{IPM}} - B_{x'} \sin \theta_{\text{IPM}} \right) \quad (6.9)$$

By substituting Eq. 6.5 and Eq. 6.6 into Eq. 6.9, we find the magnetic torque about the IPM’s axial axis as

$$\tau_{z'} = \frac{V}{\mu_0} |m||B| (\sin \theta_{\text{EPM}} \cos \theta_{\text{IPM}} - \cos \theta_{\text{EPM}} \cos \theta_z \sin \theta_{\text{IPM}}) \quad (6.10)$$

Figure 6.8 (a) IPM inclined by an angle $\theta_z$, (b) $m$ rotates about the $Z'$ axis in a circular trajectory, (c) the rotating magnetic flux density $B$ at any point along the $Z$ axis (i.e., $B_z=0$).

Eq. 6.10 indicates that the torque $\tau_{z'}$ depends on the orientation of the IPM (i.e., $\theta_{\text{IPM}}$ and $\theta_z$) and the rotation of the ASMs (i.e., $\theta_{\text{EPM}}$). $V$, $|m|$ and $|B|$ are known and the
latter varies along the Z axis, for $\theta_{\text{EPM}}=0$, as shown in Fig. 6.9. In this section, $B$ is computed with the Amperian current model presented in Section 5.2.2.

In order to analyze the effects of $\theta_{\text{IPM}}$, $\theta_{\text{EPM}}$ and $\theta_{z}$ on $\tau_{z}$, at any point along the Z axis, we firstly make the IPM’s centre to coincide with the point $x=0$, $y=0$, $z=33$ mm. At this specific point, the magnitude of the magnetic flux density is 255 mT. Since Eq. 6.3 and Eq. 6.10 are valid at any point along the Z axis, an arbitrary position for the IPM’s centre along the Z axis can be chosen to conduct this analysis. However, the selection of this specific point facilitates the collection of experimental data, as presented in subsection 6.2.3. Secondly, we let $\theta_{z}$ to take the values of 0°, 30°, 60°, 75° and 90°, and finally we depict $\tau_{z}$ as a function of $\theta_{\text{IPM}}$ and $\theta_{\text{EPM}}$ for each value of $\theta_{z}$. These results are shown in Figs. 6.10-6.12 where the results are highlighted in the red and black curves to represent $\tau_{z}$ for $\theta_{\text{IPM}}=0^\circ$ and $90^\circ$, respectively.

![Figure 6.9](image_url)  

Figure 6.9 $|B|=B_{r}=B_{\theta}$ along the Z axis and $B_{z}=0$ along the Z axis, for $\theta_{\text{EPM}}=0$, $|B|=255$ mT at $x=y=0$, $z=33$ mm.

![Figure 6.10](image_url)  

Figure 6.10 (a) Magnetic torque $\tau_{z}$ for $\theta_{z}=0^\circ$ (b) $\tau_{z}$ for $\theta_{z}=0^\circ$ and $0^\circ \leq \theta_{\text{IPM}} \leq 90^\circ$. 

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When the IPM is not tilted (i.e., $\theta_z=0^0$), a peak torque of $\tau_{mn}=8.4$ mNm is always obtained regardless of the orientation $\theta_{IPM}$ of the magnetization vector $m$ as shown in Fig. 6.10. However, this peak torque decreases as $\theta_z$ increases and also with the increments of $\theta_{IPM}$. According to the results shown in Figs. 6.11-6.12, $\theta_{IPM}=90^0$ and $270^0$ are critical angles where a minimum peak torque is obtained for any given inclination $\theta_z$. For example, a minimum peak torque of 2.2 mNm is obtained at $\theta_z=75^0$, $\theta_{IPM}=90^0$ and $\theta_{EPM}=180^0$ as shown in Fig. 6.12 (a). This peak torque decreases to 0 mNm if the IPM is tilted by $\theta_z=90^0$ and its magnetization vector $m$ is oriented at $\theta_{IPM}=90^0$ or $270^0$ (i.e., when $m$ is aligned with the Y axis) as shown in Figs. 6.12 (b)-(c). At these angles, the IPM and therefore, the slider-crank mechanism will stall. However, if $\theta_{IPM}$ remains below $75^0$, a minimum peak torque of 2.2 mNm can be guaranteed on the IPM even at an IPM’s inclination of $90^0$. This torque can still be sufficient to actuate the drug release mechanism.

![Figure 6.11 Magnetic torques $\tau_z$, for: (a) $\theta_z=30^0$ and (b) $\theta_z=60^0$.](image)
Figure 6.12 Magnetic torques $\tau_{z}$ for: (a) $\theta_z=75^0$, (b) $\theta_z=90^0$, (c) $\theta_z=90^0$ and $0^0 \leq \theta_{IPM} \leq 90^0$ (peak torque of 4.2 mNm at $\theta_{IPM}=60^0$).

A peak torque of approximately 3.5 mNm has been previously estimated as an appropriate magnitude to release drugs in capsule robots [111, 120]. Therefore, we conclude that even if the IPM is tilted by $90^0$, the external magnetic system can be rotated in a way that $\theta_{IPM}$ reaches a maximum angle of $60^0$ or $75^0$ and the transmitted torque will be approximately 4.2 or 2.2 mNm, respectively. However, if higher peak torques were required, the IPM’s centre can be moved towards the centre of the system where higher magnetic flux densities are expected. For example, a peak torque of 8.4 mNm is obtained when the IPM’s centre is placed at $z=33$ mm ($x=y=0$). A maximum peak torque can be obtained when $|B|=500$ [mT] right at the centre of the system as shown in Fig. 6.9. We also find results similar to those shown in Figs. 6.10-6.12 for $90^0<\theta_z<360^0$ due to the symmetry of the magnetic flux density. Therefore, the analysis of the magnitude of $\tau_{z}$, in the range $0^0 \leq \theta_z \leq 90^0$ is sufficient. These theoretical results are validated with the experimental results in Section 6.2.3.

6.2.3 Experimental methods

We use the same experimental setup used in Section 6.1, except we fabricated plastic angular guides that allowed us to incline the IPM with the angles of $\theta_z=30^0$, $60^0$, $75^0$ and $90^0$. The angular guides and the probe tip of the Gauss meter can be moved along the X, Y and Z axes. These displacements are controlled by the micromanipulation system constructed of XYZ stages as shown in Fig. 6.13.
Figure 6.13 Experimental setup to measure $\tau_{\psi}$ imparted by the array of ASMs; (a) angular guide mounted on the micromanipulation system, (b) angular guide to tilt the IPM by $\theta_z=30^0$ or $60^0$ (a similar guide was fabricated to allow inclinations of $75^0$ and $90^0$), (c) the IPM’s centre is placed at $x=y=0$, $z=33$ mm and tilted by $\theta_z=60^0$.

6.2.3.1 Magnetic flux density

In the first set of experiments, we measured $\mathbf{B}$ along the Z axis by fixing the external magnetic system (i.e., $\theta_{EPM}=0^0$ at all times) and only moving the tip of the probe with increments of 3 mm from -36 mm to 36 mm along the Z axis ($x=y=0$ at all times) as shown in Fig. 6.14 (a). Under these conditions, the cylindrical components of $\mathbf{B}$ can be expressed in Cartesian components as $B_x=B_r$, $B_y=B_\theta$ and $B_z=B_z$. Therefore, we measured $B_x$, $B_y$ and $B_z$ along the Z axis, although only $B_x$ is shown in Fig. 6.14 (b) because $B_y$ and $B_z$ varied between -3 mT and 3 mT. These small values of $B_y$ and $B_z$ can be taken as 0 mT for practical purposes because they will hardly contribute to the
magnetic torque transmitted to the IPM. Therefore, the static magnetic flux density \( B \) along the Z axis is equal to \( B_x \) for \( \theta_{EPM} = 0^\circ \).

Figure 6.14 (a) Experimental setup to measure \( B \) along the Z axis, (b) \( B_x \) along the Z axis and \( B_y = B_z = \pm 3 \) mT along the Z axis. \( B_x = 236 \) mT at \( x = y = 0, z = 33 \) mm.

When the external magnetic system was rotated (i.e., when \( \theta_{EPM} \) varied), \( B \) became the rotating magnetic flux density. These experimental results verified the accuracy of the analytical model for \( B \) expressed in Eq. 6.3 when the external magnetic system was rotated by an angle \( \theta_{EPM} \). We believe that the difference between the analytical and experimental results shown in Fig. 6.14 (b) (which were always less than 8.6\%) may be due to the small gaps between the ASMs and the aluminum case that holds them in place and also some small misalignments among the 24 off-the-shelf ASMs. However, these differences can decrease the magnitude of \( B \) but do not change our model expressed in Eq. 6.3.
6.2.3.2 Magnetic torque

In the second set of experiments, we firstly placed the IPM’s centre at x=y=0 and z=33 mm. This is a practical and convenient point that has allowed us to tilt the IPM up to a maximum angle of $\theta_z=90^\circ$. For a larger inclination (or if the IPM’s centre was located at points such $z<33$ mm), the plastic connector attached to the torque sensor contacted the external magnetic system and impeded the direct measurement of $\tau_{zm}$.

Similarly, points above $z=33$ mm were not of interest for two reasons: they are outside the cylindrical region of interest where the capsule robot would operate and the magnetic flux density was lower than 236 mT, which would decrease the magnetic torque to smaller values that may not be measured by our torque sensor, especially when the IPM was tilted. Secondly, we were interested in measuring $\tau_{zp}$ when the IPM’s magnetization vector was aligned with the +Y axis (i.e., $\theta_{\text{IPM}}=90^\circ$) since this is a critical angle for the transmitted torque $\tau_{zp}$ as presented in Section 6.2.2. Therefore, we conducted our second set of experiments to measure $\tau_{zp}$ as a function of $\theta_{\text{EPM}}$ and $\theta_z$ as shown in Fig. 6.15.

(a)

(b)
These experimental results validate the analytical model for $\tau_{zr}$ described by Eq. 6.10. According to these results, peak torques from 2 to 4 mNm were obtained when the IPM was tilted by an angle $\theta_z$ between $75^0$ and $60^0$, respectively. Consequently, the IPM and the crank of the drug release module can be rotated even if the IPM is tilted by these angles. However, the peak torque of $\tau_{zr}$ continued decreasing if the IPM was further inclined (i.e., for $\theta_z > 75^0$), reaching 0 mNm at $\theta_z = 90^0$. Therefore, the IPM and the crank stall for any angle $\theta_{EPM}$ if the IPM has the specific orientation determined by $\theta_{IPM} = 90^0$ and $\theta_z = 90^0$. At these values, the inclination $\phi$ (see Fig. 3.1) and perhaps the position of the external magnetic system would need to be adjusted by the clinician to activate the drug release mechanism. Depending on the need to generate different drug profiles (i.e., changes in number of doses or changes in release rates), the clinicians may be able to follow different real-time control strategies for the capsule robot by moving the external magnetic system to tailor therapeutic treatments to individuals’ needs.

6.2.3.3 The slider-crank mechanism

As shown in Fig. 6.16, we can insert the cubic IPM into its IPM case (A) which is connected to the crank of the slider-crank mechanism. We also fabricated the angular guides with orifices placed at $\theta_z = 45^0$, $75^0$ and $90^0$. These angular guides were mounted on a micromanipulation system as shown in Fig. 6.17.

![Figure 6.16](image) The disk-shaped crank is directly mounted on the IPM case (A). A: cubic IPM case, B: mobile frame with two orifices, C: the crank shaft that is aligned with the $Z'$ axis, D: platform that supports the IPM case. The origin of the system $X'Y'Z'$ coincides with the centre of the IPM.
In the final set of experiments, we were interested in observing the critical angle of inclination by which the IPM and the crank would stall. We firstly placed the IPM’s centre at x=y=0 and z=33 mm, inclined the IPM by $\theta_z=75^0$ and manually rotated the external magnetic system in the clockwise and counterclockwise directions. We observed that the IPM and the crank rotated in the same directions and never stalled regardless of the initial condition of $\theta_{IPM}$. Secondly, we moved the IPM’s centre along the Z axis towards the centre of the system (with the decrements of 10 mm) and observed that the IPM and the crank freely rotated as the external magnetic system was rotated (again irrespective of the initial condition of $\theta_{IPM}$). The same process was conducted for $\theta_z=45^0$ and the crank always rotated as it was driven by the rotation of the ASMs.

Thirdly, we placed the IPM’s centre at x=y=0 and z=33 mm but this time we inclined the IPM by $\theta_z=90^0$ and made $\theta_{IPM}=0^0$. Under these conditions, we started with the external magnetic system oriented at $\theta_{EPM}=0^0$ and manually rotated it in the counterclockwise and clockwise directions until $\theta_{EPM}$ reached approximately $30^0$ and $330^0$, respectively. We observed that the IPM and the crank rotated by approximately the same angles of $\theta_{IPM}=30^0$ and $330^0$. However, when the external magnetic system was further rotated until it reached $\theta_{EPM}=90^0$ (or $270^0$), we observed that the IPM and the crank also continued rotating until m was aligned with the Y axis (i.e., $\theta_{IPM}$ reached $90^0$ or $270^0$) but the crank stalled at these angles as we continued rotating the external magnetic system.
Finally, we moved the IPM’s centre along the Z axis (with the decrements of 10 mm) and at each point we maintained $\theta_z=90^\circ$. We observed that the crank always stalled right at $\theta_{\text{EPM}}=90^\circ$. These experimental results, which are in agreement with the theoretical results, indicate that for $\theta_z \leq 75^\circ$, the crank was driven by only the rotation of the ASMs and no adjustments in the position and/or orientation of the external magnetic system were needed. However, as the IPM and the crank were further inclined, the transmitted torque decreased to values that could no longer actuate the crank. We found that the crank stalled when the IPM was tilted by $90^\circ$ and $\mathbf{m}$ was aligned with the Y axis. These angles can be used in a real-time control strategy when it may be desired not to actuate the drug release mechanism. For example, in a clinical application, the drug release module should not be actuated when the capsule robot is still travelling to the target location. Once the capsule robot reaches its target, the clinician can adjust the position and orientation of the ASMs to activate the drug release module and generate different drug profiles by controlling the release rate, release amount and number of doses in real time.

6.3 Conclusions

We have presented the effects of changes in position and orientation of the 3.1 mm cubic IPM on the magnetic torque transmitted by an array of ASMs. In Section 6.1, we have specifically conducted the full analysis of the magnetic torque on the IPM whose centre can be located at any point within the 3-dimensional region of operation but no arbitrary orientation is allowed (i.e., its axial axis is always parallel to the Z axis of the external magnetic system). We have found that the external magnetic system is capable of actuating the IPM by simply rotating the ASMs regardless of the location of the IPM. Although the magnetic torque does decrease if the IPM’s centre moves along the Z axis, its magnitude can be increased (if needed) by adjusting the position of the ASMs so that the IPM’s centre is moved away from the Z axis.

In Section 6.2, we have analyzed the magnetic torque transmitted to the IPM that is arbitrarily orientated but its centre is restricted to move along the Z axis. We have derived an analytical model for the torque transmitted to the IPM and verified the accuracy of this analytical model with experimental results. The results have shown
that the IPM and therefore, the crank of the drug release mechanism can always be actuated if the IPM is tilted by angles lower than 75°. The crank stalled when the IPM was tilted by 90° and its magnetization vector \( \mathbf{m} \) was aligned with the Y axis. Although the actuation of the IPM and the crank were guaranteed at the maximum angular inclination of 75°, we believe that when the piston of the drug delivery mechanism is articulated with the entire mechanical system and the drug is stored in its reservoir, additional friction forces and load will be present in the capsule robot. Therefore, we expect that the maximum angular inclination of the IPM to guarantee the actuation of the drug release module should be below 75°. A localization and orientation module within the capsule robot would be very useful to indicate the clinician when this maximum angular inclination is reached. However, the integration of such a tracking module with our proposed DDS for WCE represents future work that needs to be conducted for the final medical application.

Another significant result is that although we aim to embed the IPM in a capsule robot to achieve drug delivery, the analysis of the transmitted torque on an IPM with an arbitrary position and/or orientation can be applied to any magnetic propulsion system requiring a remote transmission of a rotational magnetic field or torque to a target device. These analyses are important not only to understand the limitations of the magnetomechanical system but also to generate real-time control strategies. In Chapter 7, we extend these analyses to any point within the cylindrical region of interest rather than only at points along the Z axis.
Chapter 7
Analysis of Magnetic Torque Acting on an IPM with an Arbitrary Position and Orientation

In this chapter, we conduct the full analysis of the magnetic torque imparted to an IPM embedded in a prototype of the capsule robot for drug delivery. We allow the IPM or capsule robot to have an arbitrary position and orientation like in a typical real medical application in the GI tract. This analysis has been conducted in Chapter 6 by restricting the IPM’s position to points along the Z axis. However, in Chapter 7, we aim to analyze the rotating magnetic flux density $\mathbf{B}$ and its effects on the magnetic torque $\tau_z$, transmitted to the IPM which has an arbitrary position and orientation within a 3-D working volume. This full analysis will provide a better understanding of the magnetic coupling and be of great help for the development and achievement of real-time control of the DDS embedded in the capsule robot. The rotating $\mathbf{B}$ is generated with the optimal magnetic system made of 24 off-the-shelf ASMs shown in Fig. 5.39. The IPM is to operate within the cylindrical region defined by the internal radius of the ASMs (i.e., $r_{in} = 30$ mm) and the length of the ASMs of $\Delta z = 60$ mm. In order to facilitate the analysis of $\mathbf{B}$ and $\tau_z$, we align the general coordinate frame of $XYZ$ shown in Fig. 3.1 with the $X_aY_aZ_a$ coordinate frame of the external magnetic system shown in Fig. 3.3. We do this by making the centres of both systems to coincide so that the plane $z=0$ cuts the external magnetic system in half (i.e., cuts it at the length of $\Delta z = 30$ mm) and by putting in parallel planes $XY$ and $X_aY_a$ (i.e., the external magnetic system is not inclined, thus $\phi = 0^0$, see Fig. 3.1 for definition of $\phi$). Therefore, the external magnetic system can only rotate about the Z axis (i.e., $0^0 \leq \theta_{EPM} \leq 360^0$) to generate a rotating $\mathbf{B}$ at any point $P_i(r,\theta,z)$ within the cylindrical region of interest where the IPM’s centre would be located which is defined by $r < r_{in}$, $0^0 \leq \theta \leq 360^0$ and $-30$ mm $\leq z \leq 30$ mm.

7.1 Theory

7.1.1 Analysis of magnetic flux density

In Section 6.2, we have presented the analytical models for the rotating $\mathbf{B}$ and the transmitted torque $\tau_z$ on a tilted IPM and these models are expressed by Eq. 6.3 and Eq. 6.10, respectively. We have found that these analytical models are accurate enough to predict the maximum inclinations for the IPM with respect to the Z axis to
actuate the drug release mechanism. Beyond this maximum inclination, the position and orientation of the external magnetic system should be readjusted to provide enough magnetic torque on the tilted IPM, the centre of which is confined to points $P_i$ along the Z axis. For this region of operation, we have found that $B_z=0$ mT [130]. Therefore, $B$ rotates in planes parallel to the plane XY and its magnitude is constant as it rotates $360^\circ$, describing a circular trajectory as shown in Fig. 6.8 (c). However, for points $P_i$ outside the Z axis, the three components of $B$ vary as a function of $(r,\theta,z)$ which affect the torque transmitted to the IPM. Thus, we aim to analyze the three components of $B$ within the cylindrical region of operation of the IPM. Throughout this chapter, we use the Amperian model described in Section 5.2.2 to calculate the theoretical values of $B$. We start with the analysis of $B_z(r,\theta,z)$ to determine how it affects $B$ and $\tau_{zt}$.

The rotating $B$ created at the specific point of interest $P_i$ $(r,\theta,z)$ when the external magnetic system is rotated by an angle $\theta_{EPM}$ $(0^\circ \leq \theta_{EPM} \leq 360^\circ)$ can also be computed if we fix the external magnetic system to $\theta_{EPM}=0^\circ$ and only rotate the point of interest $P_i$ along circular trajectories (by varying $\theta$ from $0^\circ$ to $360^\circ$) and calculate the three static components of $B$ at each $P_i$. We have used this approach to derive $\tau_{zt}$ described by Eq. 6.1 and the theoretical results were verified with the experimental results presented in Section 6.1. Therefore, we use this same approach in this chapter to calculate the static $B_z(r,\theta,z)$ in the cylindrical region defined by $r \leq 27$ mm, $0^\circ \leq \theta \leq 360^\circ$, and $0 \leq z \leq 70$ mm. Although the axial region of operation of the IPM’s centre is confined to $-30$ mm $\leq z \leq 30$ mm, we have decided to analyze only the upper section of the external magnetic system (i.e., $z\geq0$) and extended it to $z=70$ mm. The theoretical results for $B_z(r,\theta,z)$ are shown in Fig. 7.1. Due to the symmetry of the external magnetic system along the Z axis, similar theoretical results are obtained for $B_z$ in the region $-70$ mm $\leq z < 0$. 
Figure 7.1 $B_z$ for: (a) z=0 mm, (b) z=10 mm (Maximum= 84 mT), (c) z=20 mm (Maximum =233 mT), (d) z=30 mm (Maximum=537 mT), (e) z=40 mm (Maximum=243 mT), (f) z=50 mm (Maximum=109 mT), (g) z=60 mm (Maximum=52 mT) and (h) z=70 mm (Maximum=25 mT). Maximum absolute values are obtained at and $\theta=0^\circ$ (same values at $180^\circ$) and with $r$ approx. at 27 mm.

Figure 7.1 indicates that $B_z(r,\theta,z)$ is non zero outside the Z axis and in the plane $z=0$. $|B_z|$ increases with the increments of $r$, reaching always maximum values when $r$ is approx. 27 mm. However, $B_z$ is approximately zero for small $r$, which means that if the capsule robot remains close to the Z axis, $B_z$ is approximately zero for any value of $\theta$ and $z$. The analytical models described by Eq. 6.3 and Eq. 6.10 can still be used for $B$ and $\tau_{zz}$, respectively. As the $r$ position of the IPM’s centre is increased, $B_z$ increases and these analytical models are no longer valid.

This indicates that if the IPM’s centre remains near the Z axis, it will be subjected to a rotating $B$ the components of which are mainly projected onto the place XY (since $B_z$ is approximately zero). As the IPM’s centre is moved away from the Z axis, $B_z$ becomes different than zero and the IPM will be subjected to a rotating $B$, the components of which are no longer confined to the plane XY. Since $|B_z|$ reaches its maximum values at $\theta=0^\circ$ (and same maximum values at $\theta=180^\circ$) for any value of $r$ and $z$, we show $B_z(r,z)$ at these two angles in Fig. 7.2. These results show that $|B_z|$ increases as $z$ increases from 0 to 30 mm and reaches its maximum values at $z=30$ mm (for any $r$ and $\theta$), and decreases again for greater values of $z$. 

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Based on the results shown in Figs. 7.1-7.2, we find that the maximum values of $B_z$ occur at $\theta=0^0$ (and also by symmetry at $\theta=180^0$) and also as $r$ and $z$ are increased. These maximum values of $B_z$ have an impact on $B$ and $\tau_{\text{mn}}$ as presented later in this chapter. For example, the magnitude of $B_z$ is maximum at the point $P_i = (r,0,z)=(26\ \text{mm},0^0,30\ \text{mm})$. Therefore, this point $P_i$ represents a critical point for the analysis of the rotating $B$ and subsequently the torque $\tau_{xy}$ transmitted to the IPM. Fig. 7.3 (a) shows the rotating $B$ (normalized, i.e., $\vec{B}$) at this point $P_i$ created when the ASMs are rotated by $\theta_{\text{EPM}}$ ($0^0 \leq \theta_{\text{EPM}} \leq 360^0$) (which is equivalent to varying $\theta$ from $0^0$ to $360^0$ and keeping $\theta_{\text{EPM}}=0^0$).
Figure 7.3 (a) Rotating $\vec{B}$ created at point $P_i=(26 \text{ mm},0^0,30 \text{ mm})$ when the ASMs are rotated $360^0$, (b) side view of $\vec{B}$ (plane rz), (c) $\vec{B}$ projected onto the plane rz, the best line $z = mr + b_0$ inclined by an angle $\varphi_{\pi}$ and $\vec{B_n}$ perpendicular to the plane $\pi$.

$\vec{B}$ is normalized and represented by $\vec{B} = \vec{B}_r u_r + \vec{B}_0 u_0 + \vec{B}_z u_z$ and the symbol $\wedge$ indicates that it is a normalized value (between 0 and 1). Fig. 7.3 (b) is a side view of the rotating $\vec{B}$ at $P_i=(26 \text{ mm},0^0,30 \text{ mm})$ which shows that $\vec{B}$ (and therefore $\vec{B}$) approximately lies in a plane $\Pi$ that is inclined by an angle $\varphi_{\pi}$ with respect to the plane $z=0$. If we project the rotating $\vec{B}$ onto the plane rz as shown in Fig. 7.3 (c) (each point generated with each change in $\theta_{EPM}$), we see that the plane $\Pi$ could be represented with a line $z = mr + b_0$ that is inclined by an angle $\varphi_{\pi}$ and $\vec{B_n}$ represents the unit vector perpendicular to the plane $\Pi$.

In our theoretical results for the rotating $\vec{B}$, we have found that as the point of analysis $P_i$ is brought closer to the Z axis (i.e., as $r$ tends to 0 mm while keeping $\theta=0^0$ and $z=30$ mm), the angle of inclination $\varphi_{\pi}$ also tends to approach zero. Furthermore, when $r=0$ mm (i.e., for points along the Z axis) and for points in the plane $z=0$, we have observed that $\varphi_{\pi}=0^0$. In these two regions, we have also found that $B_z=0$ mT as shown in Fig. 7.1. Consequently, in these two regions, $\vec{B}$ only lies in the planes parallel to the plane $z=0$ and $\vec{B_n}$ is aligned with the Z axis (i.e., $\vec{B_n}=u_z$). But outside these two regions, $B_z$ is different than 0 mT and $\vec{B}$ approximately lies in the planes $\Pi$ that are no longer parallel to the plane $z=0$. Such planes $\Pi$ are inclined by an angle $\varphi_{\pi}$ with respect to the plane $z=0$. These findings clearly indicate that there is a relationship between $B_z$ and $\varphi_{\pi}$ and that $P_i=(26 \text{ mm},0^0,30 \text{ mm})$ is a point of interest where $\varphi_{\pi}$ is maximum.
Since the plane $\theta=0^0$ (or $180^0$) is critical for the analysis of $\mathbf{B}(r,\theta,z)$ and subsequently, its effects on $\tau_z$, we present in Fig. 7.4 the changes in its three components in the plane $\theta=0^0$ for $0 \leq r \leq 26$ mm, $0 \leq z \leq 70$ mm.

Figure 7.4 Variation of the three components of $\mathbf{B}(r,0,z)$ in the plane $\theta=0^0$: (a) $B_y = B_\theta$, (b) $B_x = B_r$, (c) $B_z$. Similar results are obtained in the plane $\theta=180^0$ due to the symmetry of the external magnetic system.

According to the results shown in Fig. 7.4 (a), $B_y=0$ mT and the only components that affect the inclination $\varphi_\pi$ of the plane $\Pi$ in which the rotating $\mathbf{B}$ approximately lies are $B_r$ and $B_z$. Thus, it is sufficient to focus our attention on these two components at any point in this plane rz (i.e., $\theta=0^0$) to determine $\varphi_\pi(r,z)$. For example, at point $P=(26$ mm,$0^0,30$ mm), $B_r=201$ mT, $B_z=-478$ mT. These two values are represented by only one point along the line shown in Fig. 7.3 (c) when they are normalized. Therefore, $\varphi_\pi$ can be approximated as

$$\varphi_\pi = \tan^{-1} \left( \frac{B_z}{B_r} \right) \quad (7.1)$$
At the specific point $P_i = (26 \text{ mm}, 0^0, 30 \text{ mm})$ and using Eq. 7.1, we find $\varphi_\pi = -67.19^0$. However, $\vec{B}$ does not lie exactly in the plane $\Pi$ when the ASMs are rotated by an angle $\theta_{EPM}$ (see Figs. 7.3 (a)-(b)). Consequently, $\varphi_\pi$ and $\vec{B}_n$ can be better estimated if we use more points as depicted in Fig. 7.3 (c). We propose that a linear regression model can help us find the line $z = m r + b_0$ and with the slope $m$ we can subsequently estimate $\varphi_\pi(r,z)$ and $\vec{B}_n(r,z)$ more accurately. Once these values are found, we can use them to estimate their effects on $\tau_{zr}$ imparted to an IPM that has an arbitrary orientation. For instance, in Section 6.2, we show that for any point along the Z axis: $\vec{B}_n = u_z$, and $\vec{B}$ rotates in a plane $\Pi$ that is always parallel to the plane $z=0$ (i.e., $\varphi_\pi = 0^0$).

We have also found in Section 6.2 that the IPM stalls (i.e., $\tau_{zr} = 0 \text{ mNm}$) when there is a misalignment of $90^0$ between $\vec{B}_n$ and the axial axis $Z'$ of the IPM (i.e., when the IPM is tilted by $\theta_z = 90^0$—see Fig. 6.8 (a) for definition of $\theta_z$). Furthermore, we also have shown in Section 6.2 that $\tau_{zr}$ reaches its peak value when $\vec{B}_n$ is aligned with the axial axis of the IPM (i.e., $\theta_z = 0^0$).

Based on these results from Section 6.2, regarding the effects of the misalignment between $\vec{B}_n$ and the axial axis of the IPM on $\tau_{zr}$, we can estimate the critical inclinations for the IPM whose centre is located at point $P_i = (26 \text{ mm}, 0^0, 30 \text{ mm})$ as follows. Since we have estimated that at this point, $\varphi_\pi$ is approximately $-67^0$, then $\vec{B}_n$ has an inclination of $67^0$ with respect to the Z axis. Therefore, we estimate a maximum peak value for $\tau_{zr}$ if the axial axis of the IPM is aligned with $\vec{B}_n$ (i.e., $\theta_z = 67^0$). Similarly, we estimate $\tau_{zr} = 0 \text{ mNm}$ if there is a misalignment of $90^0$ between $\vec{B}_n$ and the axial axis $Z'$ of the IPM, which happens for the critical inclination of $\theta_z = 67^0 + 90^0 = 157^0$. Furthermore, we estimate that $\tau_{zr}$ will decrease from its peak value to $0 \text{ mNm}$ as $\theta_z$ varies from $67^0$ to $157^0$. In order to verify these results and improve their accuracy, we have used, as presented in subsection 7.1.1.1, the linear regression model to estimate $\varphi_\pi$ and in subsection 7.1.1.2, we have proposed two additional methods to estimate $\varphi_\pi$. Once $\varphi_\pi$ is obtained, we use it to analyze its effects on $\tau_{zr}$.

### 7.1.1.1 Linear regression model to estimate $\varphi_\pi$

The rotating $\vec{B}$ at point $P_i(r,\theta,z)$ is generated as the ASMs are rotated by an angle $\theta_{EPM}$, therefore each vector $\vec{B}$ is associated with a change in $\theta_{EPM}$. In order to determine the
angle of inclination \( \varphi_{\pi}(r,z) \) at any point \( P_i(r,\theta,z) \) (with \( \theta=0^\circ \) or \( 180^\circ \)), we propose the following steps:

1. Project the heads of each vector \( \vec{B} \) on the plane \( rz \) (the tail of each vector has its origin at \( P_i \)). We do this because \( B_z \) is maximum in the plane \( rz \) and also because \( \vec{B}_n \) has no tangential component (i.e., \( \vec{B}_n \cdot \vec{u}_\theta = 0 \)) as we have observed through simulations by varying \( P_i \) in the plane \( rz \) and observing that the rotating \( \vec{B} \) almost lying in the plane \( \Pi \). By assuming that \( \vec{B}_n \cdot \vec{u}_\theta = 0 \), this proposed methodology helps us prove that this assumption is valid.

2. Calculate the best line \( z = mr + b_0 \) that would go through all the projections of \( \vec{B} \) in the plane \( rz \) (using linear regression) and calculate the coefficient of determination \( R \) to measure how well the linear model predicts the data.

3. Calculate \( \varphi_{\pi} = \tan^{-1}(m) \) (\( m \) is the inclination).

4. Calculate the unit vector \( \vec{B}_n \) perpendicular to the line \( z = mr + b_0 \). This unit vector represents the optimal vector perpendicular to the plane \( \Pi \).

\[
\vec{B}_n = \cos(\varphi_c)\vec{u}_r + 0\vec{u}_\theta + \sin(\varphi_c)\vec{u}_z
\]

with

\[
\varphi_c = 90^\circ + \varphi_{\pi}
\]

5. Calculate the perpendicularity \( \alpha_{\pi} \) between the optimal \( \vec{B}_n \) (found using linear regression) and each \( \vec{B} \) as

\[
\alpha_{\pi} = \cos^{-1}(\vec{B}_n \cdot \vec{B})
\]

for each \( \vec{B} \) that is generated when the ASMs are rotated by \( \theta_{\text{EPM}} \) which varies from \( 0^\circ \) to \( 360^\circ \).

6. Since \( R \) and \( \alpha_{\pi} \) are measures of the accuracy of the linear regression model, we use these values to estimate the error incurred in the linear regression model (steps 1 to 5).

These steps are carried out, for the purpose of demonstrating how the proposed methodology works, to find \( \vec{B}_n \) at \( P_i = (r,\theta,z) = (26 \text{ mm},0^\circ,30 \text{ mm}) \) as shown in Fig. 7.3.
We find: \( z = -1.71717 r - 0.0035 \) and \( R = 0.948 \). From the slope and using Eqs. 7.2-7.3, we find \( \hat{B}_n = 0.864147 u_r + 0.003239 u_z \), with \( \phi_\pi = -59.78^0 \). Finally, we calculate \( \alpha_\pi \) using Eq. 7.4 to measure the perpendicularly of \( \hat{B}_n \) with each \( \hat{B} \) generated as the ASMs are rotated by an angle \( \theta_{EPM} \) and we present these results in Fig. 7.5. These results shows that at point \( P_i = (26 \text{ mm}, 0^0, 30 \text{ mm}) \), the rotating \( \hat{B} \) (and therefore the rotating \( B \)) can be considered to lie in the plane \( \Pi \) that is inclined by an angle \( \varphi_\pi = -59.78^0 \) with respect to the plane \( XY \). The error \( ER \) incurred in the linear model is determined by \( R \) which is approximately 1. Thus, \( ER = (1 - R)^*100\% = 5.2\% \). Similarly, the error \( E\alpha_{abs} \) due to assuming that the rotating \( \hat{B} \) perfectly lies in the plane \( \Pi \) is given by

\[
E\alpha_{abs} = \max (\alpha_{abs}) \times \frac{100\%}{90^0} \tag{7.5}
\]

with

\[
\alpha_{abs} = |90^0 - \alpha_\pi| \tag{7.6}
\]

Using Eq. 7.6 at the point \( P_i = (26 \text{ mm}, 0^0, 30 \text{ mm}) \), we find that the maximum value of \( \alpha_{abs} \) is \( 7.86^0 \) and by substituting this value into Eq. 7.5, we obtain an error \( E\alpha_{abs} \) of 8.8%.

![Graph showing changes of \( \alpha_\pi \) vs. \( \theta_{EPM} \)]

Figure 7.5 Changes of \( \alpha_\pi \) to determine the error incurred when we assume that the plane \( \pi \) contains all the vectors of the rotating \( B \).

When we use Eq. 7.1, we have estimated \( \varphi_\pi = -67.2^0 \) by considering \( B \) at the same point \( P_i \) but only for the angle \( \theta_{EPM} = 0^0 \). However, with the linear regression model
presented above, we have considered a full rotation of the ASMs (i.e., $\theta_{EPM}$ varies from $0^0$ to $360^0$ with increments of $10^0$), obtaining a more accurate estimation of $\varphi_{\pi} = -59.78^0$ with an error $E_{\alpha_{abs}}$ of 8.8%. Since this error is relatively small, it indicates that our assumption stated in step 1, namely $\vec{B}_n \cdot \vec{u}_\theta = 0$, is valid. Although these results are obtained at the specific point $P_i = (26 \text{ mm}, 0^0, 30 \text{ mm})$ to show how the proposed methodology works, we can use it along with Eqs. 7.2-7.6 to calculate $\varphi_{\pi}(r,z)$, $\vec{B}_n(r,z)$ and also the errors $E_{R}(r,z)$ and $E_{\alpha_{abs}}(r,z)$ associated with the assumption that the rotating $\mathbf{B}$ perfectly lies in the plane $\Pi$. However, before using this methodology, which we call M1, for any point $P_i$ in the plane $\theta = 0^0$, we present two additional methods to estimate $\varphi_{\pi}$ and $\vec{B}_n$ in subsection 7.1.1.2. By doing this, we compare the accuracy of the three proposed methods.

7.1.1.2 Additional methods to estimate $\varphi_{\pi}$

In this subsection, we propose another method, called M2, in order to further evaluate the efficacy of M1. M2 consists of the following steps:

1. Calculate all the prospective normal vectors $\vec{B}_n$ by finding the cross product between $\vec{B}(r, \theta_i, z)$ and $\vec{B}(r, \theta_j, z)$ where the subscripts $i$ and $j$ are used to represent different angular positions of the ASMs and $i \neq j$ (for example: if the ASMs are rotated $n$ times with increments of $360^0/n$, then $i: 1, 2, \ldots, n$ and $j=i+1, i+2, \ldots, n$).

2. Calculate $\alpha_{M2} = \cos^{-1}(\vec{B}_n \cdot \vec{B})$ for each $\mathbf{B}$ that is generated when the ASMs are rotated by $\theta_{EPM}$ from $0^0$ to $360^0$ and for all the prospective normal vectors $\vec{B}_n$. $\alpha_{M2}$ measures the perpendicularity between $\vec{B}_n$ and each $\mathbf{B}$.

3. Calculate $\alpha_{absM2} = |90^0 - \alpha_{M2}|$. $\alpha_{absM2}$ is a matrix and its number of rows equals the number of vectors $\vec{B}_n$ and its number of columns equals the number of times the ASMs are rotated (i.e., $n$).

4. Calculate the maximum value of $\alpha_{absM2}$ in each row. Subsequently, calculate the index $K$ where the minimum value of the remaining column vector occurs. $K$ is the index that indicates the optimal normal vector $\vec{B}_{n_{opt}}$ among all the prospective vectors $\vec{B}_n$ that minimizes the maximum of $\alpha_{absM2}$. In Matlab, we use the functions max and min to calculate $K$ and find $\vec{B}_{n_{opt}}$. 

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5. Calculate \( \varphi_c = \tan^{-1}\left( \frac{B_{z_{opt}}}{B_{r_{opt}}} \right) \) where \( B_{r_{opt}} \) and \( B_{z_{opt}} \) are the radial and axial components of \( \hat{\mathbf{B}}_{\text{opt}} \), respectively. This unit vector is the optimal unit vector perpendicular to the plane \( \Pi \).

6. Calculate \( \varphi_\pi = \varphi_c - 90^\circ \) (i.e., the inclination of the plane \( \Pi \) with respect to the plane \( XY \)).

These steps are carried out to find \( \hat{\mathbf{B}}_{\text{opt}} \) at \( P_i = (r, \theta, z) = (26 \text{ mm}, 0^\circ, 30 \text{ mm}) \) as shown in Fig. 7.3. All the prospective vectors \( \hat{\mathbf{B}}_n \) generated from Step 1 are shown in Fig. 7.6.

![Figure 7.6 Rotating \( \mathbf{B} \) at \( P_i = (r, \theta, z) = (26 \text{ mm}, 0^\circ, 30 \text{ mm}) \) (represented with the purple vectors) and \( \hat{\mathbf{B}}_n \) represented with the red vectors.](image)

By using M2, we have found: \( \hat{\mathbf{B}}_{\text{opt}} = 0.8623\hat{\mathbf{u}}_r + 0\hat{\mathbf{u}}_\theta + 0.5064\hat{\mathbf{u}}_z \) and \( \varphi_\pi = -59.57^\circ \).

These are similar results obtained using the linear regression method (i.e., M1). Matlab also has a built-in function called fminmax that can be used to calculate \( \hat{\mathbf{B}}_{\text{opt}} \) and \( \varphi_\pi \). We have used fminmax and called this method M3. Thus, the problem of finding \( \hat{\mathbf{B}}_{\text{opt}} \) can be described as:
$\min \max F(B_n)$ such that $\text{ceq}(B_n) = 0$

with $F(B_n) = \left\| \frac{\pi}{2} - \cos^{-1}(B_n \cdot \hat{B}) \right\|$ and $\text{ceq}(B_n) = \|B_n\| - 1$.

$\text{ceq}(B_n)$ is used to restrict the solution $B_n$ to be a unit vector and $F$ measures the perpendicularity between $B_n$ and all the vectors $\hat{B}$ generated as $\theta_{EPM}$ varies from $0^0$ to $360^0$ at the point $P_i$. By using M3, we have found $\hat{B}_{\text{nopt}} = 0.8619u_r + 0u_\theta + 0.5070u_z$ and $\varphi_{\pi} = -59.53^0$ which are similar to the results obtained from M1 and M2.

Using Eq. 7.4 and Eq. 7.6, we calculate $\alpha_{\text{abs}}$ as a function of $\theta_{EPM}$ for the three methods M1, M2 and M3 and this comparison is shown in Fig. 7.7. These results from the three methods at $P_i = (r,0,z) = (26 \text{ mm}, 0^0, 30 \text{ mm})$ demonstrate that $\hat{B}_n \cdot u_\theta = 0$ as we have assumed in the step 1 of M1. Furthermore, the comparison of the accuracy of the three methods shown in Fig. 7.7 suggests that any of them can be used to estimate $\varphi_{\pi}$ and $\hat{B}_n$. Consequently, $\alpha_{\text{abs}}$ along with Eq. 7.5 can also be used to estimate the error incurred in assuming that the rotating $\hat{B}$ lies perfectly in the plane $\Pi$.

![Figure 7.7 $\alpha_{\text{abs}}$ as a function of $\theta_{EPM}$](image)

**7.1.1.3 The inclination of the plane \( \Pi \) as a function of \((r,z)\)**

In order to determine $\varphi_{\pi}$ at any point $P_i(r,0,z)$, we have used the linear regression method (i.e., M1). Fig. 7.8 shows variations of $\varphi_{\pi} = f(r,z)$ (keeping $\theta=0^0$, that is: the plane rz) within two cylindrical regions of operation (note: the same results for $\varphi_{\pi}$ are
obtained for any $0 \leq \theta \leq 360^\circ$). However, our region of interest for the capsule robot is $0<r\leq 26$ mm, $0^\circ \leq \theta \leq 360^\circ$, $-30$ mm $\leq z \leq 30$ mm and in this region, the linear regression model is accurate enough (the maximum error is 8.8% as presented in subsection 7.1.1.1).

Figure 7.8 shows that $\varphi_{\pi} = 0^\circ$ for points $P_i$ along the Z axis (a result that was presented in Section 6.2) and also for all the points $P_i$ in the plane $z=0$. For any $P_i$ within the cylindrical region of interest ($r \leq 26$ mm, $-30$ mm $\leq z \leq 30$ mm, and $0^\circ \leq \theta \leq 360^\circ$), the inclination of the plane $\Pi$ is given by $\varphi_{\pi}$. Furthermore, we find that $\varphi_{\pi} \geq 0^\circ$ for $z \leq 0$ and $\varphi_{\pi} < 0^\circ$ for $z > 0$.

![Figure 7.8 Cylindrical region defined by: (a) 2 mm $<$ r $<$ 28 mm (steps of 2 mm), -40 mm $<$ z $<$ 40 mm (steps of 5 mm), any value of $\theta$, (b) 2 mm $<$ r $<$ 26 mm (steps of 2 mm), 5 mm $<$ z $<$ 40 mm (steps of 5 mm), any value of $\theta$.](image)

Using Eq. 7.5, we calculate the error $E_{\alpha_{\text{abs}}}(r,z)$ incurred by assuming that the rotating capsule perfectly lies in the plane $\Pi$ and the results are shown in Fig. 7.9. Similarly, the error $ER=(1-R)*100\%$ for the linear regression model, for all the points within the same cylindrical regions of operations, is shown in Fig. 7.10.
Figure 7.9 Error $E_{\alpha_{\text{abs}}}$ in the cylindrical region defined by: (a) $2 \text{ mm} \leq r \leq 26\text{ mm}$ (steps: 2 mm), $5\text{ mm} \leq z \leq 70\text{ mm}$ (steps: 5 mm), any value of $\theta$, (b) $2 \text{ mm} \leq r \leq 26\text{ mm}$ (steps: 2 mm), $5\text{ mm} \leq z \leq 30\text{ mm}$ (steps: 5 mm), any value of $\theta$.

Figure 7.10 Error $E_{\alpha_{\text{abs}}}$ in the cylindrical region defined by: (a) $2 \text{ mm} \leq r \leq 26\text{ mm}$ (steps: 2 mm), $5\text{ mm} \leq z \leq 70\text{ mm}$ (steps: 5 mm), any value of $\theta$; b) $2 \text{ mm} \leq r \leq 26\text{ mm}$ (steps: 2 mm), $5\text{ mm} \leq z \leq 30\text{ mm}$ (steps: 5 mm), any value of $\theta$.

Figure 7.9 (a) and Fig. 7.10 (a) show that at $r=26\text{ mm}$ and $z=45\text{ mm}$, the maximum errors $E_{\alpha_{\text{abs}}}$ of 59.3% and ER of 98.7% occur. Fig. 7.9 (b) and Fig. 7.10 (b) show that at $r=26\text{ mm}$ and $z=30\text{ mm}$, the maximum errors $E_{\alpha_{\text{abs}}}$ of 8.8% and ER of 5.2% are obtained. Therefore, within the cylindrical region of operation of the capsule robot, a maximum error of 8.8% obtained when using the linear regression model (i.e., M1) and it occurs at the point $P_i(r, \theta, z) = (26\text{ mm}, \theta, 30\text{ mm})$ for any value of $\theta$. Figs. 7.9-7.10 also show that M1 is not accurate for points $P_i = (26\text{ mm}, 0, 45\text{ mm})$ at which an error of
98.7% occurs due to a low value of R from the linear regression model (as shown in Fig. 7.11 (c), the rotating $\mathbf{B}$ which does not lie in a defined plane). However, the results from Figs. 7.9-7.10 do indicate that the method M1 can be used for $r \leq 20$ mm (any value $\theta$ and any value of $z$ such $-70 \leq z \leq 70$ mm) and the error is kept below 6%. Furthermore, since a relatively small error $E\alpha_{abs}$ is obtained, we conclude that $\mathbf{B}_n \cdot \mathbf{u}_\theta = 0$ in the entire cylindrical region of operation for the capsule robot as stated in the step 1 of M1. Fig. 7.11 shows the rotating $\mathbf{B}$ at the points $P_i(r,\theta,z) = (20$ mm, $0^0, 30$ mm), $(26$ mm, $0^0, 30$ mm), $(26$ mm, $0^0, 45$ mm), respectively.

![Figure 7.11](image-url)

According to the results for $E\alpha_{abs}(r,z)$, this error increases with $r$ and $z$, reaching a maximum value of 8.8% at point $P_i(r,\theta,z) = (26$ mm, $0^0, 30$ mm). Therefore, in our
experimental section (Section 7.2), we measure the magnetic flux density $\mathbf{B}$ and the magnetic torque $\tau_z$, for different points in the plane $z=30$ mm.

Before carrying out experiments, we have also used Comsol to compare the numerical results for the rotating $\mathbf{B}$ with the analytical results at the point of interest $P_i(r,\theta,z) = (26$ mm, $0^0, 30$ mm). Fig.7.12 shows the external magnetic system along with the circular trajectory, which is shown in blue colour, with a radius of 26 mm in the plane $z=30$ mm. To compute the three components of $\mathbf{B}$ at the fixed point $P_i$ as the external magnetic system rotates $360^0$ (i.e., $0^0 \leq \theta_{EPM} \leq 360^0$) is equivalent to computing the three components of $\mathbf{B}$ along the circular trajectory and fix $\theta_{EPM} = 0^0$ at all times. This latter approach is used in Comsol and Figs. 7.13-7.14 show the comparison of the three components of the rotating $\mathbf{B}$. In these results, the subscript $m$ is used to indicate that the results computed using Matlab (using the Amperian models described in Section 5.2.2) while the subscript $c$ is used to indicate that the result is obtained using Comsol. For instance, $B_{rm}$ and $B_{rc}$ are the radial components of $\mathbf{B}$ computed using Matlab and Comsol, respectively. The results obtained with Comsol validate our analytical results for the angle of inclination $\phi_\Pi$ of the plane $\Pi$ in which $\mathbf{B}$ approximately lies and also the unit vector $\hat{\mathbf{B}}_n$ perpendicular to the plane $\Pi$.

Figure 7.12 Model of the external magnetic system in Comsol and the circular trajectory with a radius $r$ of 26 mm in the plane $z=30$ mm.
7.1.1.4 Analysis of the magnitude of the rotating magnetic flux density $B$

In subsections 7.1.1.1-7.1.1.3, we have analyzed the rotating $B$ and the plane $\Pi$ in which the rotating $B$ approximately lies and we have also estimated how the angle of inclination $\varphi_\pi$ of the plane $\Pi$ can affect the magnetic torque $\tau_{zr}$. In this subsection, we analyse the magnitude of the rotating flux density $B$ (i.e., $|B|$) at any point $P_i(r, \theta, z)$ within the cylindrical region of operation of the capsule robot. This analysis is important
because the transmitted torque depends on \( \mathbf{B} \), therefore understanding \( \mathbf{B} \) gives us a better understanding of the transmitted torque.

In Section 6.2, we have shown that the rotating \( |\mathbf{B}| \) at any point \( P_i \) along the Z axis, is described by a circle as the ASMs are rotated by an angle \( \theta_{\text{EPM}} \). This has facilitated the analysis of the effects of \( \mathbf{B} \) on \( \tau_{xy} \), because \( \mathbf{B} \) can be easily expressed in Eq. 6.3 and subsequently an analytical model for \( \tau_{xy} \), which is expressed by Eq. 6.10, has been derived. However, for points \( P_i \) outside the Z axis, the rotating \( |\mathbf{B}| \) is no longer described by a circle and therefore its effects on \( \tau_{xy} \) are different. In order to quantify how \( |\mathbf{B}| \) changes at any point \( P_i (r, \theta, z) \) as a function of \( \theta_{\text{EPM}} \), we can project \( \mathbf{B} \) onto the plane \( \Pi \), which we call \( \mathbf{B}_\Pi \) as shown in Fig. 7.15, and then show the magnitude of this projected vector (i.e., \( |\mathbf{B}_\Pi| \approx |\mathbf{B}| \)) as \( \theta_{\text{EPM}} \) varies from 0° to 360°. \( \mathbf{B}_\Pi \) can be derived as follows:

\( \mathbf{B} \) at any point \( P_i \) is expressed by Eq. 5.1, using the general cylindrical coordinate system \((\mathbf{u}_r, \mathbf{u}_\theta, \mathbf{u}_z)\). \( \mathbf{B} \) can also be expressed in a new coordinate system using the unit vectors \((\mathbf{u}_{r1}, \mathbf{u}_{\theta1}, \mathbf{u}_{z1})\) as

\[
\mathbf{B}(r, \theta, z) = B_{r1}(r, \theta, z)\mathbf{u}_{r1} + B_{\theta1}(r, \theta, z)\mathbf{u}_{\theta1} + B_{z1}(r, \theta, z)\mathbf{u}_{z1} \quad [T] \quad (7.7)
\]

where the unit vectors \( \mathbf{u}_{r1} \) and \( \mathbf{u}_{\theta1} \) are an orthogonal base for the plane \( \Pi \) on which \( \mathbf{B} \) approximately lies. We can choose any orthogonal base to generate the plane \( \Pi \), however we have decided to choose a convenient orthogonal base to facilitate our analysis of \( |\mathbf{B}| \) as shown in Fig. 7.15. In this convenient orthogonal base, \( \mathbf{u}_{r1} \) is aligned with the line \( z = mr + b_0 \) and its direction is determined by the angle \( \varphi_\pi \). We also choose, for the orthogonal base, \( \mathbf{u}_{\theta1} \) to be aligned with \( \mathbf{u}_\theta \). Finally, \( \mathbf{u}_{z1} \) is aligned with \( \mathbf{B}_n \) which is perpendicular to the plane \( \Pi \) as shown in Fig. 7.3 (c).

Since a relatively small error \( E_{\alpha_{\text{abs}}} \) is incurred in assuming that \( \mathbf{B} \) perfectly lies in the plane \( \Pi \) as presented in subsection 7.1.1.3, then we can neglect \( B_{z1} \) in Eq. 7.7. Therefore, \( \mathbf{B} \) is approximately equal to \( \mathbf{B}_\Pi \) which is given by

\[
\mathbf{B} \approx \mathbf{B}_\Pi = B_{r1}\mathbf{u}_{r1} + B_{\theta1}\mathbf{u}_{\theta1} \quad (7.8)
\]

We are interested in finding \( B_{r1} \) and \( B_{\theta1} \) which are the two components of \( \mathbf{B} \) projected onto the plane \( \Pi \). Multiplying Eq. 5.1 by \( \mathbf{u}_{r1} \) (dot product):
Figure 7.15 The rotating $\mathbf{B}_\Pi$ at point $P_i$ lies in the plane $\pi$ that is spanned by the unit vectors $\mathbf{u}_{r1}$ and $\mathbf{u}_{\theta 1}$.

$$\mathbf{B} \cdot \mathbf{u}_{r1} = B_{r1} = B_r \mathbf{u}_r \cdot \mathbf{u}_{r1} + B_\theta \mathbf{u}_\theta \cdot \mathbf{u}_{r1} + B_z \mathbf{u}_z \cdot \mathbf{u}_{r1} \quad (7.9)$$

Since $\mathbf{u}_\theta \cdot \mathbf{u}_{r1} = 0$ (because $\mathbf{u}_{\theta 1} = \mathbf{u}_\theta$), $\mathbf{u}_r \cdot \mathbf{u}_{r1} = \cos (\varphi_\pi)$, and $\mathbf{u}_z \cdot \mathbf{u}_{r1} = \cos(90^\circ - \varphi_\pi) = \sin (\varphi_\pi)$ (for any value of $\varphi_\pi$), therefore $B_{r1}$ in Eq. 7.9, for any value of $\varphi_\pi$, is reduced to

$$B_{r1} = B_r \cos (\varphi_\pi) + B_z \sin(\varphi_\pi) \quad (7.10)$$

Similarly, we multiply Eq. 5.1 by $\mathbf{u}_{\theta 1}$ and find $B_{\theta 1} = B_\theta$. Following the same procedure, we multiply Eq. 5.1 by $\mathbf{u}_{z1}$ and find $B_{z1} = -B_r \sin (\varphi_\pi) + B_z \cos(\varphi_\pi)$, although this component is neglected and not used when calculating $\mathbf{B}_\Pi$. Finally, we substitute $B_{\theta 1}$ and Eq. 7.10 into Eq. 7.8, and find that $\mathbf{B}_\Pi$ can be described, at any point $P_i$, as a linear combination of the vectors $\mathbf{u}_{r1}$ and $\mathbf{u}_{\theta 1}$ as follows:

$$\mathbf{B} \approx \mathbf{B}_\Pi = (B_r \cos (\varphi_\pi) + B_z \sin(\varphi_\pi)) \mathbf{u}_{r1} + B_\theta \mathbf{u}_{\theta 1} \quad (7.11)$$

For example, at the specific $P_i (r=26 \text{ mm}, \theta=0^\circ, z=25 \text{ mm})$, we calculate $\varphi_\pi$ (using the method M1 described in Section 7.1.1.1) and substitute it into Eq. 7.11 and obtain how $|\mathbf{B}_\Pi|$ rotates in the plane $\Pi$ as shown in Figs. 7.16-7.17.
At this point $P_i$, the magnitude of $B_\|\|$ does not resemble an ellipse or a circle as presented in Section 6.2. If $|B_\|\|$ was described by a circle, Fig. 7.17 would show a constant value (as it is predicted by Eq. 6.3 for a rotating magnetic flux density that is described by a circle).

According to the results in Figs. 7.16-7.17, the maximum of $|B_\|\|$ is 691.8 mT and it occurs at $\theta_{EPM\text{max}} = 60^\circ$ (and by symmetry also at $120^\circ$, $240^\circ$ and $300^\circ$). Similarly, the minimum of $|B_\|\|$ is 420.3 mT and it occurs at $\theta_{EPM\text{min}} = 90^\circ$ (and by symmetry also at $270^\circ$).
$270^\circ$. $\theta_{\text{EPMmax}}$ and $\theta_{\text{EPMmin}}$ are important angles to be considered in the transmitted torque. For example, if the IPM’s centre is located at this point $P_i$ and its axial axis is tilted by $\theta_z = |\varphi_\Pi|$ (i.e., the magnetization vector of the IPM $\mathbf{m}$ lying in the plane $\Pi$), then a maximum magnetic torque can be imparted to the IPM if the ASMs are rotated $\theta_{\text{EPMmax}}$ and $\mathbf{m}$ is misaligned by $90^\circ$ with respect to $\mathbf{B}_\Pi$.

We have also used Eq. 7.11 for all points $P_i$ within the cylindrical region of operation of the capsule robot and have found that as $r$ is closer to 0 (for any $-30 \text{ mm} < z < 30 \text{ mm}$ and any $\theta$), the magnitude of $\mathbf{B}_\Pi$ (and subsequently $|\mathbf{B}|$) does not vary much but remains almost constant. Therefore, the shape of the rotating $|\mathbf{B}_\Pi|$ is more like a circle. This is because the inclination of the plane $\Pi$, $\varphi_\Pi$, tends to be $0^\circ$ as $r$ approaches the Z axis. Thus, Eq. 7.11 becomes Eq. 5.1 which can be expressed as Eq. 6.3 when the ASMs are rotated. These results indicate that the magnitude of the magnetic flux density is described by a circle only when $\mathbf{B}_n$ is aligned with the Z axis but the circular shape is degenerated as the misalignment between $\mathbf{B}_n$ and the Z axis is increased.

### 7.1.2 Analysis of the axial magnetic torque

In this subsection, we aim to analyze the axial magnetic torque $\tau_{zr}$ by deriving an analytical model for $\tau_{zr}$. This torque is imparted to an IPM that has an arbitrary position and orientation within the cylindrical region of operation (i.e., at any point $P_i$). Due to the symmetry of the magnetic flux density (as it has been presented in subection 7.1.1), the analysis of $\tau_{zr}$ for points such $0 \leq z \leq 30 \text{ mm}$ is sufficient. The orientation of the IPM is fully determined by the parameters ($\theta_{\text{IPM}}, \theta_z$) as it was presented in Section 6.2. Since $\mathbf{m}$ rotates about the axial axis $Z'$ of the IPM (i.e., in the plane $X'Y'$) in a circular trajectory as shown in Fig. 6.8 (b), then $\mathbf{m}$ can be expressed by Eq. 6.2. However, $\mathbf{m}$ can also be described as a linear combination of the unit vectors ($\mathbf{u}_r$, $\mathbf{u}_\theta$, $\mathbf{u}_z$) which span the plane $X'Y'$. Therefore, ($\mathbf{u}_{xr}$, $\mathbf{u}_{yr}$) and ($\mathbf{u}_{r'}$, $\mathbf{u}_{\theta}$) are interchangeable in Eq. 6.2. In this section (Section 7.1.2), we represent $\mathbf{m}$ as a function of ($\mathbf{u}_{r'}$, $\mathbf{u}_{\theta}$) using Eq. 6.2 and align $\mathbf{u}_\theta$, with $\mathbf{u}_\theta$ as we have done in Section 6.2.

To find an expression for $\tau_{zr}$ at any point $P_i$, we use the 2-D vector representation shown in Fig. 7.18 in which ($\mathbf{u}_r$, $\mathbf{u}_\theta$, $\mathbf{u}_z$) is the orthogonal base for the general coordinate system $XYZ$. In this vector representation, the 3-D plane $\Pi$ (spanned by the
unit vectors $\mathbf{u}_{r1}$ and $\mathbf{u}_{01}$ as shown in Fig. 7.15 in which $\mathbf{u}_{01}$ is aligned with $\mathbf{u}_0$) is reduced to the line directed by $\mathbf{u}_{r1}$ and inclined by an angle $\varphi_{\pi}$ with respect to $\mathbf{u}_r$. $\mathbf{u}_{z1}$ is aligned with $\mathbf{B}_n$ and it is perpendicular to $\mathbf{u}_{r1}$. Furthermore, the axial axis $Z'$ of the IPM is directed by the unit vector $\mathbf{u}_z$, that forms an angle $\theta_z$ with respect to $\mathbf{u}_z$. In this section, we vary $\theta_z$ in the range $0^0 \leq \theta_z \leq 180^0$. Therefore, in this 2-D vector representation $\mathbf{u}_0 = \mathbf{u}_{01} = \mathbf{u}_0$, and all the analysis for $\tau_{z}$ can be carried out in the plane $rz$ (i.e., any plane $\theta$) as shown in Fig. 7.18. We choose the plane $\theta$=0$^0$ and 0$\leq z$$\leq$30 mm for our analysis.

Figure 7.18 2-D vector representation to derive an equation for $\tau_{zr}$.

The analytical model for $\tau_{z}$ at any point $P_i$ can be derived by following a similar process used in Section 6.2. Since $\mathbf{m}$ rotates in the plane $X'Y'$ spanned by the unit vectors $(\mathbf{u}_{r'}, \mathbf{u}_{\theta'})$, we can simply project the rotating $\mathbf{B}$ onto the same plane $X'Y'$ and then use Eq. 3.1 and take only the axial component of $\mathbf{r}$ (i.e., $\tau_{z}$). In subsection 7.1.1.4, we have shown that $\mathbf{B}$ can be approximately projected onto the plane $\Pi$ by $\mathbf{B}_\Pi$ which is given in Eq. 7.8 as a linear combination of the unit vectors $(\mathbf{u}_{r1}, \mathbf{u}_{01})$. However, $\mathbf{B}_\Pi$ can also be described as a linear combination of $(\mathbf{u}_r, \mathbf{u}_\theta, \mathbf{u}_z)$ as follows

$$\mathbf{B}_\Pi = B_{r'}, \mathbf{u}_{r'} + B_{\theta'}, \mathbf{u}_{\theta'} + B_{z'}, \mathbf{u}_{z'}$$

(7.12)

Therefore, Eq. 7.12 is an analytical expression of $\mathbf{B}$ projected (approximately) onto the plane $X'Y'$. By substituting $\mathbf{m}$ and $\mathbf{B}$ (expressed in Eq. 7.12) into Eq. 3.1 and taking only the axial component of $\mathbf{r}$, we find

$$\tau_{z'} = \frac{V}{\mu_0} |\mathbf{m}| \left( B_{\theta'}, \sin \theta_{\text{IPM}} - B_{r'}, \cos \theta_{\text{IPM}} \right) \quad \text{[Nm]}$$

(7.13)
Equation 7.13 shows that only the components $B_r$ and $B_\theta$, contribute to $\tau_{zr}$. In order to find the expressions of these components as a function of $(B_r, B_\theta, B_z)$ and the angles $\varphi_\pi$ and $\theta_z$, we proceed as follows. Firstly, by definition $u_\theta = u_{\theta'} = u_{\theta_1}$, therefore, we find $B_\theta = B = B_\theta$. Secondly, we multiply (dot product) Eq. 7.8 by $u_r$, to find an expression for $B\{n$ and $B\approx n$ contribute to $\tau_{mn}$. In order to find the expressions of these components as a function of $(B\{n, B\approx n, & B_m)$ and the angles $\varphi_\pi$ and $\theta_m$, we proceed as follows. Firstly, by definition $\sqrt{f} = \sqrt{f_{\approx}} = \sqrt{f_{\sim}}$, therefore, we find $B\approx n = B\approx = B\approx$. Secondly, we multiply (dot product) Eq. 7.8 by $\sqrt{\sum n}$ to find an expression for $B\{n$. But $u_{\theta_1} \cdot u_r = 0$ (because $u_{\theta_1} = u_{\theta'}$), and $u_r \cdot u_r = \cos (\varphi_\pi + \theta_z)$, therefore Eq. 7.14 reduces to

$$B_r = B_{r_1} \cos (\varphi_\pi + \theta_z)$$

(7.15)

Thirdly, we substitute Eq. 7.10 into Eq. 7.15 and find

$$B_r = (B_r \cos (\varphi_\pi) + B_z \sin (\varphi_\pi)) \cos (\varphi_\pi + \theta_z)$$

(7.16)

Finally, we substitute Eq. 7.16 and $B_\theta = B$ into Eq. 7.13 and find

$$\tau_{z'} = \frac{V}{\mu_0} |m| (B_\theta \cos \theta_{\text{IPM}} - [(B_r \cos (\varphi_\pi) + B_z \sin (\varphi_\pi)) \cos (\varphi_\pi + \theta_z)] \sin \theta_{\text{IPM}}) \quad \text{[Nm]}$$

(7.17)

Equation 7.17 is valid for any point $P_i(r, \theta, z)$ within the cylindrical region of operation of the capsule robot. The three components of $B$ (i.e., $B_r, B_\theta$, and $B_z$) vary at each point $P_i$ and also with $\theta_{\text{EPM}}$ as the external magnetic system rotates. Furthermore, $\varphi_\pi$ varies at each point $P_i$, as it has been presented in subsection 7.1.1.3. Consequently, $\tau_{z'}$ is a function of the position $P_i$, the orientation of the IPM (given by $\theta_{\text{IPM}}$ and $\theta_z$) and the angle of rotation $\theta_{\text{EPM}}$ of the ASMs. Eq. 17 reduces to Eq. 6.10 for any point $P_i$ along the $Z$ axis.

In order to analyze the effects of the IPM's position and orientation and $\theta_{\text{EPM}}$ on $\tau_{zr}$, we firstly choose the point $P_i(26 \text{ mm}, \theta = 0^0, 33 \text{ mm})$. At this point $\varphi_\pi$ is approximately $-60^0$ (see subsections 7.1.1.1.7.1.1.2). Secondly, we vary $\theta_z$ from $0^0$ to $180^0$ (with increments of $10^0$), and finally we simulate $\tau_{zr}$ using Eq. 7.17 as a function of $\theta_{\text{IPM}}$ and $\theta_{\text{EPM}}$ for each value of $\theta_z$. We have observed that $\tau_{zr}$ has similar trends to those shown.
in Figs. 6.10-6.12. We find that $\tau_z$ is 0 mNm for any value of $\theta_{EPM}$ when the IPM is orientated by $\theta_{IPM}=90^0$ (i.e., when $m$ is aligned with $u_a$) and $\theta_z=150^0$ as shown in Fig. 7.19. Because $\theta_{IPM}=90^0$ (or $270^0$) is a critical angle for the IPM’s magnetization vector, we have also decided to fix $\theta_{IPM}=90^0$ and vary $\theta_z$ and $\theta_{EPM}$ and the result for $\tau_z$, is shown in Fig. 7.20.

Figure 7.19 $\tau_z$, for $\theta_z=150^0$, $P_i(r,0,z)=(26 \text{ mm},0^0,30 \text{ mm})$.

Figure 7.20 $\tau_z$, for $0^0 \leq \theta_z \leq 180^0$, $\theta_{IPM} = 90^0$, $P_i(r,0,z) = (26 \text{ mm},0^0,30 \text{ mm})$. Peak torque of approx. 18 mNm is reached when $\theta_z=60^0$. 
According to these results for \( \tau_z \), obtained at the specific point \( P_i(r,0,z) = (26 \text{ mm}, 0^0, 30 \text{ mm}) \), the IPM (and subsequently the crank) stalls at the critical IPM’s orientation, which we denote with the subscript \( c \), given by \( \theta_{zc} = 150^0 \) and \( \theta_{IPMC} = 90^0 \) (or \( 270^0 \)). \( \theta_{zc} \) and \( \theta_{IPMC} \) are the zeros of Eq. 7.17 for any value of \( \theta_{EPM} \). We also find the same critical angles at any point \( P_i(r,0,z) = (26 \text{ mm}, 0, 30 \text{ mm}) \), for any value of \( \theta \) (i.e., the circular trajectory shown in Fig. 7.12). Furthermore, we obtain the following relationship between \( \theta_{zc} \) and \( \varphi_{\pi} \):

\[
\theta_{zc} = 90^0 - \varphi_{\pi}(r,z)
\]  

(7.18)

This relationship indicates that \( \tau_z = 0 \) mNm (for any value of \( \theta_{EPM} \)) if there is a misalignment of \( 90^0 \) between the axial axis \( Z' \) of the IPM and \( \vec{B}_n \) and \( \theta_{IPM} \) reaches \( \theta_{IPMC} = 90^0 \) (or \( 270^0 \)). Further, according to Fig. 7.20, a maximum peak torque of approximately 18 mNm is obtained when the IPM is tilted by \( \theta_z = \theta_{zm} = |\varphi_{\pi}| \) (i.e., when there is no misalignment between the axial axis \( Z' \) of the IPM and \( \vec{B}_n \) which means that \( m \) and \( B \) rotate in the same plane).

We have also extended our simulations for \( \tau_z \) to any point within the cylindrical region of interest. By finding the zeros of Eq. 7.17 for points in the plane \( rz \), regardless of the values of \( \theta \) and \( \theta_{EPM} \), we obtain the results for the critical orientations \( \theta_{zc} \) and \( \theta_{IPMC} \) where the IPM stalls as shown in Fig. 7.21s (a)-(b), respectively. In our simulations: \( \theta_z \) is varied with increments of \( 2^0 \), \( 2 \text{ mm} < r < 28 \text{ mm} \) (increments of 2 mm), \( 0 \text{ mm} < z < 30 \text{ mm} \) (increments of 5 mm), any value of \( \theta \). We have also used Eq. 7.18 to obtain indirectly \( \varphi_{\pi} \) from \( \theta_{zc} \) and this is shown in Fig. 7.21 (c).
These results indicate that the IPM stalls for the critical angles $\theta_{zc}$ (given by Eq. 7.18) and $\theta_{IPMC}=90^0$ (or 270$^0$) at any point $P_i$. We have also found that the inclination of the IPM where a peak torque is obtained is indeed $\theta_z = \theta_{zm}=\vert\varphi_{\pi}\vert$. Therefore, the misalignment angle between the axial axis $Z'$ of the IPM and $\hat{B}_n$ determines how much torque is imparted to the IPM: obtaining a peak torque when there is no misalignment and decreasing to zero mNm when the misalignment reaches 90$^0$ and $\theta_{IPMC}=90^0$ (or 270$^0$). For example, at any point in the circular trajectory $P_i (r=26 \text{ mm, } \theta,z=30 \text{ mm})$, we have found that $\theta_{zm}=\vert\varphi_{\pi}\vert=60^0$ and the peak torque is approximately 18 mNm (see Fig. 7.20). We have also found that the torque becomes zero mNm when $\theta_{zd}=150^0$ (and $\theta_{IPM} = 90^0$ or 270$^0$). If $\theta_z$ is increased from $\theta_{zm}$ to $\theta_{zc}$, the peak torque can decrease to values that can no longer actuate the slider-crank mechanism. In Section 6.2, we have found that a peak torque of approx. 2 mNm (which is sufficient to actuate the crank mechanism) is guaranteed $\theta_z \leq 75^0$ (for points in the Z axis). However, for points in the circular trajectory $P_i (r=26 \text{ mm, } \theta,z=30 \text{ mm})$, this angle of inclination of the IPM’s axial axis is increased to almost 82$^0$ (i.e., $\theta_z-\theta_{zm}=142^0-60^0$) as shown in Fig. 7.22 where the peak torques of 3 mNm are reached at $\theta_z=140^0$. This increment of the maximum angle between the axial axis $Z'$ of the IPM and $\hat{B}_n$ from $\theta_z=75^0$ to 82$^0$ is due to an increase in the magnitude of $\mathbf{B}$ in those points that are near the ASMs. This can be seen by comparing the results of $\mathbf{B}$ shown in Fig. 6.9 and Fig. 7.17.

Figure 7.21 the zeros of Eq. 7.17: (a) $\theta_{zc}$, (b) $\theta_{IPMC}$, (c) $\varphi_{\pi}$ obtained indirectly from Eq. 7.18.
According to these results presented in this section, the analysis of the rotating magnetic field for any point \( P_i \) is sufficient to estimate \( \varphi_{\Pi} \). Once \( \varphi_{\Pi} \) is calculated, we can simply calculate \( \theta_{zm} = |\varphi_{\Pi}| \) and then use Eq. 7.18 to estimate \( \theta_{zc} \). In Section 7.2, we conduct experiments for the magnetic flux density and the magnetic torque to validate our theoretical results.

7.2 Experimental Results

A 3-channel Gauss meter (Lakeshore-Model 460) was used to measure the magnetic flux density generated by the ASMs and a torque gauge (HTG2-40 supplied by IMADA) with its respective torque sensor was used to measure the axial magnetic torque on the 3.1 mm cubic IPM. The probe tip of the Gauss meter can be moved along the X and Z axes and the arrays of magnets can only be moved along the Y axis. These displacements are controlled by a micromanipulation system based on an X-Y-Z stage, as shown in Fig. 6.2.

7.2.1 Experiments for magnetic flux density

We measured the three components of the rotating \( \mathbf{B} \) at \( P(r, \theta = 0^\circ, z = 30 \text{ mm}) \) for \( r=14 \text{ mm} \) and \( r=22 \text{ mm} \) as we rotated the ASMs by an angle \( \theta_{EPM} \) from \( 0^\circ \) to \( 360^\circ \) with the increments of \( 15^\circ \). These experimental results are compared with the theoretical results from the Amperian model (i.e., Model2 presented in Section 5.2.2) and are shown in Figs. 7.23-7.25.
Figure 7.23 Comparison of the radial components of the rotating $B$ for: (a) $r=14$ mm, (b) $r=22$ mm.
Figure 7.24 Comparison of the tangential components of the rotating $B$ for: (a) $r=14$ mm, (b) $r=22$ mm.

Figure 7.25 Comparison of the axial components of the rotating $B$ for $r=14$ mm and $r=22$ mm.

According to the results presented in Figs. 7.23-7.25, there is a good agreement between the experimental and the theoretical results. Therefore, our experimental results validate our theoretical results obtained using the methodology described in Subsection 7.1.1.1 and Comsol for the angle of inclination $\varphi_\pi$ of the plane $\Pi$, at any point $P_i$, on which the rotating $B$ lies. Furthermore, these experimental results also validate $\mathbf{B}_n$ (the vector perpendicular to the plane $\Pi$) expressed in Eq. 7.2.
7.2.2 Experiments for the magnetic torque

In our experiments to measure directly \( \tau_{zr} \), we placed the IPM’s centre at \( P_i(r = 20 \text{ mm}, \theta = 0^\circ, z = 30 \text{ mm}) \). This is a practical and convenient point that has allowed us to tilt the IPM up to a maximum angle of \( \theta_z=80^0 \) with respect to the Z axis as shown in Fig. 7.26. For a larger inclination (or if the IPM’s centre was located at points such \( z<30 \text{ mm} \)), the plastic connector attached to the torque sensor contacted the external magnetic system and impeded the direct measurement of \( \tau_{zr} \) (similar to the reason why we chose \( z=33 \text{ mm} \) to measure directly \( \tau_{zr} \) in subsection 6.2.3.2). We also aligned the IPM’s magnetization vector \( \mathbf{m} \) with the +Y axis (i.e., \( \theta_{\text{IPM}}=90^0 \)) because this is a critical angle for \( \tau_{zr} \). Therefore, we let \( \theta_z \) to take the values of \( 0^0, 45^0, \) and \( 135^0 \) and completed a full rotation of the ASMs (with increments \( \Delta \theta_{\text{EPM}}=15^0 \)) at each value of \( \theta_z \). The comparison of theoretical and experimental results for the torque \( \tau_{zr} \), as a function of \( \theta_z \) and \( \theta_{\text{EPM}} \) is shown in Fig. 7.27.

![Image of experiments](a) (b) (c)

Figure 7.26 The IPM was tilted by approx.: (a) \( \theta_z=0^0 \), (b) \( \theta_z=45^0 \), and (c) \( \theta_z=135^0=-45^0 \).
Figure 7.27 Comparison of theoretical and experimental results for the axial torque \( \tau_z \) on the IPM that was tilted by: (a) \( \theta_z = 0^0 \) and (b) \( \theta_z = 45^0 \). When \( \theta_z = 135^0 \), we measured approx. \( \tau_z = 0 \) mNm for any value of \( \theta_{EPM} \) and for this reason this result is not presented in this figure.

The experimental results in Fig. 7.27 show that at \( P_i \), and when \( \theta_{EPM} = 180^0 \), the peak torque changed from 8 mNm to a peak torque of 12 mNm and then decreased to 0 mNm when the IPM was inclined by \( \theta_z = 0^0 \), \( \theta_z = 45^0 \), and \( \theta_z = 135^0 \), respectively. Thus, experimentally \( \theta_z = \theta_{zm} = 45^0 \) (and consequently we estimate experimentally \( \phi_n = -45^0 \)). Furthermore, the experimental results presented in Fig. 7.27 indicate that when the IPM’s centre is located at \( P_i (r = 20 \text{ mm}, \theta = 0^0, z = 30 \text{ mm}) \), and with the specific orientation determined by \( \theta_{IPM} = 90^0 \) (or \( 270^0 \)) and \( \theta_z = \theta_{zc} = 135^0 \) (or \( -45^0 \)), the magnetic torque \( \tau_z \) is approximately 0 mNm regardless of the values of \( \theta_{EPM} \). This result validates Eqs. 7.17-7.18 and under these circumstances the IPM and (subsequently) the slider-crank mechanism would stall.

These experimental results for \( \theta_{zm} = 45^0 \theta_{zc} = 135^0 \) are also predicted by the analysis of the rotating \( \mathbf{B} \) at \( P_i \) which indicates that \( \mathbf{B} \) lies approximately in the plane \( \Pi \) that is inclined by an angle \( \phi_n = -46^0 \). From the analysis of \( \mathbf{B} \), we find that the angle of \( \mathbf{B}_n \) (the unit vector normal to the plane \( \Pi \)) with respect to the Z axis is \( |\phi_n| = 46^0 \) which is approx. the same as \( \theta_{zm} = 45^0 \). In order to find the critical orientation of the IPM (where it stalls) by merely analysing \( \mathbf{B} \), we only need to use Eq. 7.18 (i.e., subtract \( \phi_n \) from
90° which gives approx. 135°). We can do this because a misalignment of 90° between \( \mathbf{B}_n \) and the axial axis of the IPM makes the IPM to stall.

### 7.2.3 Experiments for the crank mechanism

We fabricated a prototype of the crank mechanism that was articulated with a 3.1 mm cubic IPM which was inserted into its case (A). The platform between the IPM and the crank was also connected to a cylindrical frame (B) that has an external diameter of 18 mm and a length of 5 mm as shown in Fig. 7.28.

![Figure 7.28 A prototype of the crank mechanism: A: cubic IPM case, B: cylindrical frame.](image)

In our first set of experiments, we placed the IPM’s centre at approximately \( P_i(r = 16 \text{ mm}, \theta = 0^\circ, z = 30 \text{ mm}) \). At this point, the theoretical analysis of \( B \) predicts \( \varphi_n = -37^\circ \). By using Eq. 7.18, we predict theoretically that the IPM would stall at \( \theta_z = 90^\circ - \varphi_n = 127^\circ \) when \( \theta_{IPMC} = 90^\circ \). Therefore, in our first experiments, we tilted the IPM by approximately \( \theta_z = 127^\circ \) (see Fig. 7.29 (a)) and rotated the ASMs. We observed that indeed the IPM and the crank stalled at the predicted critical angles \( \theta_z \) and \( \theta_{IPMC} \). In our second experiments and without changing \( P_i \), we tilted the IPM by approximately \( \theta_z = 90^\circ \) (as shown in Fig. 7.29 (b)) and observed that the IPM and the crank were successfully rotated as the ASMs rotated.
In our second set of experiments, we inserted the prototype of the crank mechanism into a transparent tube with the IPM’s magnetization vector $\mathbf{m}$ aligned with the Y axis, the IPM was tilted by $\theta_z=90^0$ and the IPM’s centre was approximately located at $P_i(r = 0 \text{ mm}, \theta = 0^0, z = 30 \text{ mm})$ as shown in Fig. 7.29 (c). We observed that the IPM and the crank stalled as the ASMs were rotated because the IPM was orientated at its critical angles $\theta_{zc}$ and $\theta_{IPMC}$.

These experiments validate our theoretical models for the angle of inclination $\varphi_\pi$ of the plane $\Pi$, at any point $P_i$, on which the rotating $\mathbf{B}$ lies. Furthermore, these experimental results also validate $\mathbf{B}_n$ (the vector perpendicular to the plane $\Pi$) expressed in Eq. 7.2, the analytical model for $\tau_{\pi r}$ expressed in Eq. 7.17 and the critical angles of orientation of the IPM where it stalls (i.e., Eq. 7.18). Eq. 7.18 can be used, for example, in a real-time control strategy because $\theta_{zc}$ is a function of $(r, z)$ (i.e., the
position of the centre of the IPM). Once the position and orientation of the capsule robot are known by means of a tracking method, the position and orientation of the ASMs can be adjusted to accurately change \( \tau_z \), as needed so that the drug release module can fully be controllable in terms of release rate, release amount and number of doses.

### 7.2.4 Experiments for the drug release mechanism embedded in a capsule robot

We fabricated with a 3D printer three different prototypes of capsule robots (Prt1, Prt2 and Prt3) as shown in Figs. 7.30-7.32 to test the capability of the magnetomechanical system to release different drug payloads including water and sunscreen. The prototypes Prt1 and Prt3, (shown in Fig. 7.30 and Fig. 7.32, respectively) were fabricated with a slider-crank mechanism in which a platform is placed between the IPM and the crank as the design shown in Fig. 7.28. This platform was moved to the bottom of the IPM in the prototype Prt3 shown in Fig. 7.31 to reduce the friction between the IPM and the crank. In all these prototypes, we embedded a 3.1 mm cubic IPM (N50).

The drug reservoir volume \( v_{dds} \) can be calculated as:

\[
v_{dds} = \pi \left( \frac{\varnothing_p}{2} \right)^2 \frac{2R}{1000} \quad [\text{mL}] \tag{7.19}
\]

with \( \varnothing_p \) as the diameter of the piston, and \( R \) is the crank length of the slider-crank mechanism. In the three prototypes, we fabricated with \( \varnothing_p = 13 \) mm and \( R = 3 \) mm. Thus, \( v_{dds} \) is approx. 0.8 mL which is an adequate volume for a DDS in WCE as presented in Table 2.1 (see column named Drug Reservoir Volume) and in Section 3.1.
Figure 7.30 Prototype Ptr1 with a crank and connecting rod lengths of 3 mm and 15 mm, respectively. External diameter of 20 mm and total length of 30 mm.

Figure 7.31 Prototype Ptr2 with a crank and connecting rod lengths of 3 mm and 10 mm, respectively. External diameter of 15 mm and total length of 30 mm. A: drug reservoir, B: the IPM articulated with the slider-crank mechanism, C: cover. (a) the three parts A, B and C assembled, (b) the disassembled capsule robot, (c) the IPM connected to the slider-crank mechanism and (d) the dimensions of the cylindrical piston.
We tested the prototypes Ptr2 and Ptr3 in transparent tubes and glass containers filled with water to simulate a more realistic environment as shown in Fig. 7.33. In our experiments, when the prototypes were tested in the glass containers that were filled with water, we glued the capsules to the bottom to avoid the rotation of the entire capsule. However, when we tested the prototypes of capsule robots in the transparent tubes, there was no need to glue the capsule to the surface of the plastic tube and the drug release module was successfully actuated. The glass containers offered larger volumes of operation to the prototypes than the more restricted volume of operation offered by the plastic tube. These tests suggest that depending on the geometry of the environment the capsule robot may not need an anchoring system to release drugs. Furthermore, the intestine tissues are deformable and could oppose a limited force to magnetic dragging which may help to stabilize the capsule robot as the DDS is actuated.

Although the gastrointestinal tract is more complex than the tube and the glass container in terms of geometry and ability to deform, we conducted our experiments under such environments only to test the capability of the prototypes to release payloads by changing the position and orientation of the capsule robots. Our results clearly show that the drug release mechanism can be activated by only rotating the ASMs in the clockwise or anticlockwise direction. Different drug profiles can be generated by adjusting the relative position and orientation of the ASMs with respect to the capsule robot. The ability to generate different drug profiles (i.e., to fully control the release amount, release rate and number of doses) gives the flexibility needed by
clinicians to tailor therapeutic treatments to individuals’ needs. Although the quantification of the capability to generate different drug profiles was not conducted in this study, we suggest that this can be carried out as future work by following a similar procedure conducted in a recent study in DDS for capsule robots [131].

![Image](a)

![Image](b)

![Image](c)

Figure 7.33 (a) Ptr2 in transparent tube, (b) Ptr3 in a transparent tube, (c) Ptr3 in a glass container filled with water.

### 7.3 Conclusions

We have analyzed the rotating magnetic field generated by an external magnetic system in the cylindrical region of operation of the capsule robot. We have found, by means of analytical models and FEM solutions in Comsol, that the magnetic field rotates in approximately planes that are directed by its normal vector $\mathbf{B}_n$ and this vector varies its direction at each point in the region of operation. Three different
methods have been used to estimate $\mathbf{B}_n$ including a linear regression model that is also useful to estimate the maximum error incurred in our theoretical models which we have found to be 8.8%. The theoretical analysis also indicates that the magnitude of the magnetic flux density is described by a circle only when $\mathbf{B}_n$ is aligned with the Z axis but the circular shape is degenerated as the misalignment between $\mathbf{B}_n$ and the Z axis is increased. Both factors (the direction of $\mathbf{B}_n$ and the magnitude of the magnetic flux density) have their own effects on the magnetic torque $\tau_z$, transmitted to the IPM that has an arbitrary position and orientation.

The magnetic torque can be fully controlled by adjusting the relative position and orientation of the external magnetic system with respect to the coordinate system of the IPM (capsule robot). The magnetic torque becomes zero mNm when the IPM’s magnetization vector is aligned with the Y axis and the axial axis of the IPM has a misalignment of $90^\circ$ with respect to $\mathbf{B}_n$. On the other hand, the maximum peak torque is transmitted when there is no misalignment between the axial axis of the IPM and $\mathbf{B}_n$. The full analysis of the effects of arbitrary position and orientation of the capsule robot on the transmitted torque to the IPM has helped us to find approximate analytical models that would enable us to carry on real-time control strategies. However, for the implementation of real-time control strategies, the loop should be closed by means of, for example, a compatible tracking system. Thus, the full control of the DDS requires a tracking system to close the loop so that once the capsule’s position and orientation are known, it would be possible to transmit magnetic torques more accurately than what it would be possible in an open-loop control.

Our theoretical models and analyses for the rotating magnetic field and the magnetic torque have been compared with the experimental results. These experimental results, which are in agreement with our theoretical results, validate our anaytical models for the rotating magnetic field and the transmitted torque to the IPM that is arbitrarily oriented at any position within the region of operation of the capsule robot. We fabricated different prototpyes of capsule robots with a drug release module that we tested to verify the capability of the magnetic actuation system to control the release rate, release amount and number of doses. These experiments with prototypes of capsule robots, which were placed in vynil tubes and glass containers filled with water,
demonstrate the ability of the magnetomechanical system to actuate the drug release module at arbitrary positions and orientations. If there is no need to activate the drug release mechanism, the external magnetic system can be adjusted to guarantee a misalignment of $90^0$ between the axial axis of the IPM and $\mathbf{B}_n$. 
Chapter 8
Conclusions and Recommendations for Future Research

This thesis has investigated the establishment of an active drug delivery system (DDS) to be embedded in the next generation of capsule endoscopes. A fully controllable DDS is a minimally invasive medical device that can help clinicians perform therapeutical procedures in the gastrointestinal (GI) tract. To this aim, a magnetomechanical system for an active drug delivery module in capsule robots is proposed. Substantial theoretical and experimental work has been conducted to investigate the feasibility and optimisation of the magnetic actuation systems for the activation and controllability of the drug release module embedded in the prototypes of robotic endoscopic capsules.

8.1 Conclusions

The following conclusions are drawn from the results presented in this thesis:

- A torque-driven magnetic device is proposed for the development of a drug release module embedded in a capsule robot. The magnetic torque and the rotating magnetic field are not affected when the external magnetic system is scaled up and the operating distance is simultaneously increased. Therefore, the torque-driven magnetic devices, as proposed in this thesis, are more suitable for WCE than force-driven magnetic devices.

- Our drug release module can generate different drug profiles by allowing the control of the release rate, release amount and number of doses. This controllability can be achieved by adjusting the relative position and orientation between the external magnetic system and the capsule robot.

- The optimization of the magnetic linkage allows the increase of the operating distance (i.e., the relative distance between the external magnetic system and the capsule robot) and the miniaturization of the on-board permanent magnet in the capsule robot.
Both outcomes (larger operating distances and minimum volume of the DDS), which have been the technical limitations in the literature, are overcome by the magnetomechanical system proposed in this thesis. The magnetic linkage is optimized by improving both the external magnetic system and the embedded permanent magnet within the capsule robot.

The external magnetic system made of permanent magnets can be optimized in terms of its design, shape, configuration and dimensions to enhance the magnetic field required to operate the capsule robot. All these optimization methodologies have been conducted by means of accurate analytical models, FEM solutions (Comsol) which were experimentally validated.

The arc-shaped permanent magnets (ASMs) are the optimal shapes to be used in our proposed external magnetic system. An optimal design and configuration of ASMs with optimal dimensions has been found and fabricated.

For the real medical application of DDS for WCE, the external magnetic system can be made of off-the-shelf permanent magnets. Our size optimization methodology demonstrates that the external magnetic system can be fabricated with a minimum volume to generate an adequate magnetic field to activate the drug release module. Therefore, this size optimization eases the maneuverability of the external magnetic system.

The external magnetic system can be fabricated with radially magnetized ASMs or tangentially magnetized ASMs or a combination of them. A magnetic structure made of only radially magnetized ASMs generates higher magnetic fields than tangentially magnetized ASMs. However, an optimal external magnetic system is obtained when both types of ASMs are used with optimal dimensions.

The shape and size of the on-board permanent magnet can also be optimized. Our results indicate that, for the same volume, cylindrical permanent magnets perform better than cubic permanent magnets under the same external magnetic field. Therefore, higher magnetic torques are imparted to the drug
delivery system based on the slider-crank mechanism. Thus, more emphasis has been placed on the optimization of the external magnetic system to improve the magnetic linkage between the internal magnet and the external magnetic system.

- Our results indicate that there is an optimal region for the actuation of the drug release mechanism. This region consists of all the points in the plane \( z=0 \) and as the radial distance of the capsule robot comes closer to the internal surface of the external magnetic system (i.e., as it comes closer to the internal boundary). In this optimal region, maximum peak torques are transmitted to a permanent magnet embedded in the capsule robot.

- Outside the optimal region of operation, the magnetic torque decreases as the capsule robot moves axially away from the plane \( z=0 \) and also as its radial distance decreases to zero (i.e., at radial points further away from the ASMs). However, with our optimal external magnetic system, adequate peak torques of approx. 8.5 mNm can be transmitted to activate the drug release mechanism even if the capsule robot is located at the critical extreme positions in the \( Z \) axis (i.e., at \( r=0 \) and \( z=\pm 30 \) mm).

- The on-board permanent magnet can be located at any position and with an arbitrary orientation within the entire region of operation of the capsule robot. This flexibility is due to the capability of the external magnetic system to generate rotating magnetic fields that are adequate to activate the drug release mechanism.

- The control of the relative position and orientation between the external magnetic system and the capsule robot can allow the generation of different real-time control strategies. Thus, the slider-crank mechanism that is articulated with the on-board permanent magnet can be switched on and off and the magnetic torque can be further manipulated by simply controlling the position and orientation of the external magnetic system.
8.2 Recommendations for Future Research

Additional work needs to be carried out not only to deepen the research in this field of magnetic actuation for drug delivery in WCE, but also to transfer this technology to minimally invasive tools for medical applications and their commercialization. Some of further studies are:

- The mechanical system can be optimized in its design to make it more compact. This can help reduce the total volume it uses within the capsule robot. For example, a scotch yoke mechanism or a cam mechanism can potentially improve the compactness of the drug delivery system.

- The mechanical parts can be manufactured and assembled using a more accurate technology than 3D printing. The improvement in the fabrication of the slider-crank mechanism can also help to reduce its volume.

- The mechanical parts should be designed and fabricated in materials that can withstand the piston force and torques transmitted to the on-board permanent magnet.

- The dynamics and kinematics of the on-board mechanical system needs to be evaluated at different rotational frequencies of the external magnetic system. This would help determine the limitations and time response of the drug release mechanism. These are important variables to be considered for the range of the release rates.

- The drug reservoir needs to be designed and fabricated to allow the storage of different drug compounds. This reservoir has to be properly sealed to isolate it from the rest of the components of the capsule robot.

- Additional modules such as an anchoring mechanism and an active locomotion system are important for the improvement of the targeted drug delivery based on the WCE. Furthermore, a tracking system needs to be implemented to close the loop and allow the actuation of the DDS more accurately.
• These additional modules have to be compatible with the drug release mechanism proposed in this thesis. The need of these modules also emphasizes again the requirement of miniaturizing the drug release module as much as possible.

• The drug delivery system proposed in this thesis should work together with the existing screening module that includes a camera in the WCE. This screening module would allow the clinician to release the drug payload at the precise location.

• In vitro and in vivo trials are needed for the entire capsule robot that includes a DDS with the screening module.

• A scaled up external magnetic system made of off-the-shelf permanent magnets must be fabricated along with its own mobile platform. This platform should allow the control of the position and orientation of the external magnetic system from a joystick.

• The specifications and technical details of the motors will depend on several requirements, including the weight of the EPMs and the rotational speed needed. Furthermore, if we completely shift the actuation problem from the capsule robot to the exterior of the patient’s body (which can be obtained if the IPM’s volume is shrunk but the EPMs’ dimensions and weight are increased), then we would expect to have more demands on the motors and possibly problems with the stability of the external platform. On the other hand, if we increase the IPM’s size, the demands on the motors will be less but there will be very limited volume for the integration of additional modules that are needed within the capsule robot. The actuation problem can be shifted from one extreme to the other and a point in between should be found to fulfill strict requirements. Nevertheless, the optimization of the external magnetic system (in terms of shape, configuration and dimensions as presented in Chapters 4 and 5) will ease the maneuverability of the EPMs and certainly decrease the demand on the motors. Therefore, future work with optimal scaled up EPMs
and a capsule robot with multiple on-board modules needs to be conducted to establish the specific technical requirements for the motors.
References


