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Development of a Proprioceptive Sensing for Soft Continuum Robots

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Declaration of Originality

I, Amirhosein Alian, declare that the work presented in this thesis is entirely my own, and the contributions of others have been duly acknowledged. A complete list of references is provided in the bibliography.

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I would like to end this section as a note to myself by quoting a few lines from the book *“The Boy, The Mole, The Fox and The Horse”*. Upon reaching a dense, dark jungle on their way back home, the boy said, “I cannot see a way through”. The horse asked, “Can you see your next step?” “Yes,” The boy replied. “Just take that,” said the horse.

Abstract

Over recent decades, the incidence of early-onset **Gastrointestinal (GI)** cancer has risen considerably, highlighting the importance of regular screening. Conventional endoscopy is inherently invasive, requiring both patient sedation and highly skilled operators. Robotic flexible endoscopy offers enhanced precision, manoeuvrability, and procedural efficiency. Soft endoscopes are promising due to their dexterity and ability to adapt to the complex curvilinear geometry of the **GI** tract.

The intrinsic nonlinearity of soft robots poses significant challenges for control, primarily due to the lack of compatible state estimation mechanisms. Existing sensing technologies suffer from limitations in scalability, integrability, cost, and biocompatibility, particularly for endoscopic applications. This thesis explored the development and validation of an **Electrical Impedance Tomography (EIT)**-based sensing methodology for **Soft Fluidic Actuators (SFAs)**, aimed at advancing proprioception and exteroception for lower **GI** soft continuum endoscopes.

Initially, the concept was validated through simulations and planar experiments. A **Flexible Printed Circuit (FPC)** array was embedded into a **Soft Continuum Manipulator (SCM)**, allowing distributed impedance measurements. A **Long Short-Term Memory (LSTM)** model reconstructed the manipulator's planar tip displacement, confirming the feasibility of data-driven shape estimation.

The **EIT** framework was extended to exteroception through tactile interaction studies. A dual-mode sensing approach enabled the estimation of contact magnitude, with further translation into soft tissue phantom palpation.

In **3D**, **EIT**, through a stretchable **FPC**, and a **Multi-Layer Perceptron (MLP)** network achieved an average **Root Mean Square Error (RMSE)** of 0.37 mm. Further validation in clinical scenarios achieved an **RMSE** of 1.34 mm under *ex vivo* conditions, followed by an *in vivo* trial in a 35 kg porcine model, where usability, robustness, and safety were evaluated.

Overall, this work established **EIT** as a compact, scalable, and biocompatible perception for soft continuum endoscopes. The findings provided a foundation for future research in transfer learning and **EIT**-based closed-loop control.

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Acronyms

1D One Dimensional.

2D Two Dimensional.

3D Three Dimensional.

ANN Artificial Neural Network.

ASD Average Standard Deviation.

CAD Computer-Aided Design.

CNN Convolutional Neural Network.

cPDMS conductive Polydimethylsiloxane.

CVs Configuration Variables.

DDCL Deep Direct Cascade Learning.

DNN Deep Neural Network.

DOF Degrees of Freedom.

EIS Electrochemical Impedance Spectroscopy.

EIT Electrical Impedance Tomography.

EKF Extended Kalman Filter.

EM Electromagnetic.

EMT Electromagnetic Tracking.

EPD Endoscopic Pneumatic Dilatation.

ESD Endoscopic Submucosal Dissection.

FBG Fibre Bragg Grating.

FCNN Fully Connected Neural Network.

FDM Frequency Division Multiplexing.

FEA Finite element analysis.

FEM Finite Element Method.

FPC Flexible Printed Circuit.

GI Gastrointestinal.

HKSTP Hong Kong Science and Technology Parks.

IACUC Institutional Animal Care and Use Committee.

IMU Inertial Measurement Unit.

LSTM Long Short-Term Memory.

MAE Mean Absolute Error.

MIS Minimally Invasive Surgery.

ML Machine Learning.

MLP Multi-Layer Perceptron.

MRC Multi-scale Medical Robotics Centre.

MRI Magnetic Resonance Imaging.

MSE Mean Squared Error.

NN Neural Network.

NURBA non-uniform rational B-spline.

PCC Piecewise Constant Curvature.

PCHIP Piecewise Cubic Hermite Interpolating Polynomial.

PLA Polylactic Acid.

POEM Per-Oral Endoscopic Myotomy.

PREM Per-Rectal Endoscopic Myotomy.

RBF-NN Radial Basis Function Neural Network.

RMSE Root Mean Square Error.

RNN Recurrent Neural Network.

SCC Soft Continuum Colonoscope.

SCM Soft Continuum Manipulator.

SD Standard Deviation.

SFA Soft Fluidic Actuator.

SNR signal-to-noise ratio.

SOFA Simulation Open Framework Architecture.

TCN Temporal Convolutional Network.

TDM Time-Division Multiplexing.

VAE Variational Autoencoder.

List of Publications

Conference Papers

Alian, A., Zari, E., Avery, J. P., Franco, E., Runciman, M., and Mylonas, G. Shape Sensing for a Soft Continuum Manipulator Using Electrical Impedance Tomography (EIT). *Hamlyn Symposium on Medical Robotics*, 2022.

Alian, A., Zari, E., Avery, J. P., Franco, E., Runciman, M., and Mylonas, G. Electrical Impedance Tomography Sensing for Soft Continuum Surgical Manipulators. *The 11th Conference on New Technologies for Computer and Robot Assisted Surgery*, 2022.

Alian, A., Mylonas, G., and Avery, J. Soft Continuum Actuator Tip Position and Contact Force Prediction Using Electrical Impedance Tomography and Recurrent Neural Networks. In *2023 IEEE International Conference on Soft Robotics (RoboSoft)*, pp. 1–6. IEEE, April 2023.

Alian, A., Avery, J. P., and Mylonas, G. Proprioceptive Shape Reconstruction of Fluidic Actuators for Endoscopic Applications. *The 17th Hamlyn Symposium on Medical Robotics*, 2025.

Alian, A., Avery, J. P., and Mylonas, G. Clinical Assessment of Soft Endoscope Shape Reconstruction Using Electrical Impedance Tomography. *The 14th Conference on New Technologies for Computer and Robot Assisted Surgery*, 2025.

Journal Papers

Alian, A., Zari, E., Wang, Z., Franco, E., Avery, J. P., Runciman, M., and Mylonas, G. Current Engineering Developments for Robotic Systems in Flexible Endoscopy. *Techniques and Innovations in Gastrointestinal Endoscopy*, vol. 25, no. 1, 2023, pp. 67–81.

Alian, A., Avery, J., and Mylonas, G. Tissue Palpation in Endoscopy Using EIT and Soft

Actuators. *Frontiers in Robotics and AI*, vol. 11, 2024, article 1372936.

Alian, A., Avery, J. P., and Mylonas, G. Electrical Impedance Tomography and Neural Networks for Shape Sensing in Soft Continuum Endoscopic Robots. *IEEE Robotics and Automation Letters* (Under Review), 2025.

Chapter 1

Introduction

1.1 Thesis Overview

As surgery increasingly progresses towards autonomy and minimal invasiveness, it is essential to focus on technologies that fulfil these requirements through compliance and self-awareness of shape and morphology. Consider colonoscopy, for instance, where visibility is limited and autonomy is crucial. The rising incidence of colorectal cancer, coupled with the increasing number of early-onset cases, necessitates regular endoscopic screenings and therapeutic interventions [1]. Given that colonoscopy is inherently complex and demands a steep learning curve, this heightened demand imposes a significant workload on both novice and experienced gastroenterologists [2, 3].

The development of advanced technologies to enable safe and reliable colon navigation can help alleviate the clinical and operational burden associated with endoscopic procedures. One effective approach is the implementation of closed-loop control strategies for both navigation and tissue manipulation. Closed-loop control facilitates safe movement and precise tip positioning, which is particularly valuable in procedures such as [Endoscopic Submucosal Dissection \(ESD\)](#) [4]. State feedback mechanisms, including shape reconstruction and tip localisation, play a critical role in supporting autonomous navigation and instrument control. Additionally, the introduction of shape sensing methods providing real-time visualisation of the endoscope's configuration has proven effective in reducing complications such as perforation and looping [5].

Furthermore, an effective way to encourage patients to undergo regular colonoscopies is to minimise the invasiveness and discomfort associated with the procedure. Although colonoscopy is inherently minimally invasive, it often remains a painful and uncom-

fortable experience for patients [6]. This issue can be mitigated by introducing tissue-compatible morphologies and technologies [7]. In this regard, the emerging field of soft robotics offers significant potential, enabling the development of soft and flexible colonoscopes designed to reduce tissue trauma and improve overall safety during endoscopic procedures.

The hyperelastic morphology of soft robots necessitates advanced control systems in conjunction with accurate, non-linear dynamic models to achieve desirable performance. One approach to facilitating the control of such systems is the incorporation of robust and reliable state estimation modalities that are consistent with the compliant nature of soft robots. This can enhance control performance by providing feedback while simultaneously reducing the computational complexity associated with analytical modelling. The application of soft robots in clinical settings imposes additional constraints on the selection of suitable state estimation technologies. Existing sensing modalities in soft robotics remain contentious for clinical deployment and integration, largely due to limitations in cost-effectiveness, biocompatibility, and immunity to [Electromagnetic \(EM\)](#) interference. Furthermore, an effective sensing modality must preserve the structural integrity of the soft robot, enable seamless integration within the instrument, and support further miniaturisation.

This thesis aims to investigate the development of an innovative sensing system based on [Electrical Impedance Tomography \(EIT\)](#), which was previously implemented for the deformation detection of a type of variable stiffness soft robots [8]. The proposed method detects shape variations reflected by a conductive medium, which is the same medium used for actively deforming the robot. This dual functionality promotes scalability and miniaturisation by enabling embedded sensing hardware with minimal size, achieved through reduced wiring and compact circuitry. As a result, additional free space is made available for the integration of surgical instruments such as an *iKnife* and biopsy forceps, which are essential for procedures such as [ESD](#).

The [EIT](#) approach provides a low-profile mechanical sensing system that preserves the structural integrity of the continuum instrument. Compared to the state of the art, this method is low cost while simultaneously offering advantages inherent to alternative modalities, including [EM](#) immunity and biocompatibility. These capabilities align with the requirements of medical applications and enable proprioception in state-of-the-art colonoscopes composed of [Soft Fluidic Actuators \(SFAs\)](#).

The proposed sensing system is transferable to other clinical applications, provided that they are composed of [SFAs](#), and its applicability extends well beyond colonoscopy. By

utilising [EIT](#) and [Machine Learning \(ML\)](#) techniques, the system estimates the endoscope's pose by measuring the impedance changes caused by deformations. These estimated shapes can then be used to determine the position of the end-effector tip, offering precise localisation and backbone reconstruction of surgical instruments, thereby enhancing the autonomy of endoscopes [9]. After demonstrating the feasibility of [EIT](#) for recognising deformations of an [SFA](#) with in-plane bending, the applicability of [EIT](#) in free motion and constrained motion under clinical settings such as *ex vivo* and *in vivo* is further explored.

Furthermore, the capabilities of the proposed sensing method extend beyond proprioception, as it holds promise for adaptation to tactile sensation and exteroception. In colonoscopy, vision-based diagnostics often struggle with early-stage cancer detection, particularly for non-polypoid lesions, which necessitates the use of soft tactile sensors for effective diagnosis [10]. Early malignant tissues typically exhibit mechanical properties, such as increased stiffness, that differ from those of healthy tissues, highlighting the importance of tactile sensing for tissue palpation [11]. This stiffness variation is reflected in the shape of the endoscope during tissue probing [12]. The proposed shape-sensing method can distinguish these variations by reconstructing the endoscope's shape, thereby enabling tissue palpation without the need for additional sensing modalities.

1.1.1 Thesis Contribution

This thesis begins with a clinical background, highlighted in Chapter 1, underscoring the significance of regular [GI](#) diagnostics, with particular emphasis on the incidence of colorectal cancer. This chapter explores the sensing modalities developed for continuum robots, specifically focusing on surgical technologies and procedures related to the lower [GI](#) tract. This chapter is partially based on a paper [13] first authored by the author of this thesis.

In Chapter 2, the functionality and performance of [EIT](#) sensing in detecting the deformation of soft bending actuators are examined. The fabrication of a hydraulically driven soft robot capable of in-plane bending is detailed, establishing a foundation for the empirical examination of [EIT](#) integration. Furthermore, the chapter explores the reliability and functionality of [EIT](#) in training a [Neural Network \(NN\)](#) model for estimating the tip position of the soft robot. The author presented the results of this chapter in [14].

Chapter 3 adapts the application of [EIT](#) in detecting deformations for the development of tactile sensing, enabling the differentiation of contact with tissue samples of varying stiffness. This chapter emphasises the effectiveness of [EIT](#) for tissue palpation and the

diagnosis of stiffened early-stage cancer tissues. It begins by demonstrating the concept of tip force estimation using a force transducer to measure and simulate tip force contacts. The experiments assessing the reliability of this approach and analysing the influence of the number of measurement channels are discussed in detail. The paper demonstrating the latest results of the investigations on tip force estimation was presented in [14], while [15] highlights the tactile sensing capabilities of EIT and the results of preliminary studies on this topic.

Chapter 4 introduces the EIT integration process into a fibre-reinforced manipulator featuring three hydraulic chambers capable of Three Dimensional (3D) motion. Further adjustments are explored to adapt the soft manipulator for *ex vivo* and *in vivo* trials. Key considerations include integrating an endoscopic camera and tethering the manipulator with an over-tube of appropriate length and flexibility. The data collection process, which involves recording positional and EIT measurements during the sequential pressurisation of each chamber, is elaborated upon later in this chapter. The EIT is characterised in greater detail, and subsequently, the tip localisation and shape reconstruction in free motion are outlined. The author has submitted the results of this chapter for publication, which are currently under review.

The aim of Chapter 5 is to transfer the intelligence gained in free motion and non-clinical environments, as discussed in Chapter 4, to surgical practice. Chapter 5 outlines the results obtained from clinical studies conducted at the Multi-scale Medical Robotics Centre (MRC) in Hong Kong between February and July 2024. The clinical experiments begin with *ex vivo* trials using a porcine large intestine and the soft endoscope described in the previous chapter. Data collection during interactions between the soft endoscope and colon tissue is performed and subsequently analysed to develop a robust NN model. This chapter also covers *in vivo* experiments, detailing the hardware setup specifications and the use of a live porcine subject. These trials, conducted under the supervision of a gastroenterologist, involve sampling EIT and shape data for further analysis. The *ex vivo* results are presented and will be submitted for publication in the future.

Finally, Chapter 6 highlights the discussion and conclusion regarding the development of EIT for shape sensing in soft continuum endoscopic robots.

In summary, the key contributions of this thesis are as follows:

- Investigating the characteristics of EIT sensing, focusing on robustness, repeatability, and reliability.
- Reconstructing the shape of an SFA using EIT in both planar and spatial free mo-

tion, enabling distal tip localisation and backbone pose estimation.

- Verifying the force estimation and tactile sensing capabilities of the proposed sensing method through experimental validation.
- Designing an experimental platform to evaluate the functionality of the proposed sensing modality in clinical practice.
- Demonstrating the functionality and applicability of the proposed approach through *ex vivo* and *in vivo* animal trials.

1.2 Clinical Background

Laparoscopic surgery, or **Minimally Invasive Surgery (MIS)**, has revolutionised modern surgical practice and represents one of the most significant advancements in medicine. Since its introduction in the early 1990s, **MIS** has gained widespread adoption and is now regarded as a gold-standard technique among surgeons [16]. **MIS** established a new paradigm by enabling surgical interventions in deep anatomical regions through small incisions rather than open surgery, thereby reducing intraoperative and postoperative complications, shortening operating times, and improving patient outcomes [17].

However, laparoscopic procedures are associated with a steep learning curve [18], primarily due to the fulcrum effect, wherein the surgeon’s hand movements are inverted relative to the motion of the instrument’s distal tip. The advent of robot-assisted **MIS** mitigated these challenges by providing intuitive and natural control of laparoscopic instruments, enhanced through augmented vision and microscale triangulation [19]. One class of **MIS** techniques, categorised by their mode of intervention, involves intraluminal or endoluminal applications, where treatments are delivered through an endoscopic device [20]. These systems are capable of active navigation within the lumen, equipped with steerable end effectors and integrated cameras that provide direct visual feedback from the surgical site.

To access target regions through tortuous, narrow, and elongated tracts such as those within the upper and lower **Gastrointestinal (GI)** tracts or respiratory systems, flexible endoscopic devices serve as essential instruments. In conventional and standard practice, endoscopes are manually controlled by clinicians, who steer the distal tip via a control dial mechanism. These devices typically include multiple channels, in addition to the embedded camera, for illumination, irrigation, insufflation, and tissue manipulation. As the instrument advances through the tract, the endoscopic camera provides visual feedback enabling the clinician to guide the distal end towards the tract centre until reaching the

region of interest. Among the various endoscopic procedures, colonoscopy remains one of the most prominent applications. In this procedure, the device is inserted through the anus to enable both diagnostic screening and therapeutic interventions within the colon.

1.2.1 Colonoscopy: Clinical Motivations and Challenges

Statistics regarding GI cancer, and particularly colorectal cancer, indicate an increase in cancer incidence among people aged below 50, over the past four decades [21] (see Fig. 1.1). In the US, the number of early-onset cancer cases is approximately 10.5% of the total new cases [22]. Many factors could contribute to this unprecedented rise, such as dietary habits, molecular and genetic profiles, and sedentary lifestyle [23]. Apart from primary prevention strategies like gene sequencing, regular screening colonoscopy of the asymptomatic population can be effective [1]. The benefit of performing colonoscopy has been shown by observing a drop in the mortality rate of colorectal cancer from 2015 to 2019 [24]. Enhanced therapeutic colonoscopy has also contributed to this result. Almost all GI diseases originate from the mucosa, the superficial layer of the intestine. For this reason, screening colonoscopy is essential, as it can detect cancerous neoplasms at their early stage, when it is still possible to treat them locally. A novel, powerful treatment is ESD, a surgical technique that allows en-bloc resection of early GI tumors. ESD can be executed with standard colonoscopes, but it is a technically challenging procedure that requires experienced colonoscopists. Other complex therapeutic procedures performed by colonoscopic technologies are Per-Oral Endoscopic Myotomy (POEM), Endoscopic Pneumatic Dilatation (EPD), Per-Rectal Endoscopic Myotomy (PREM) [25].

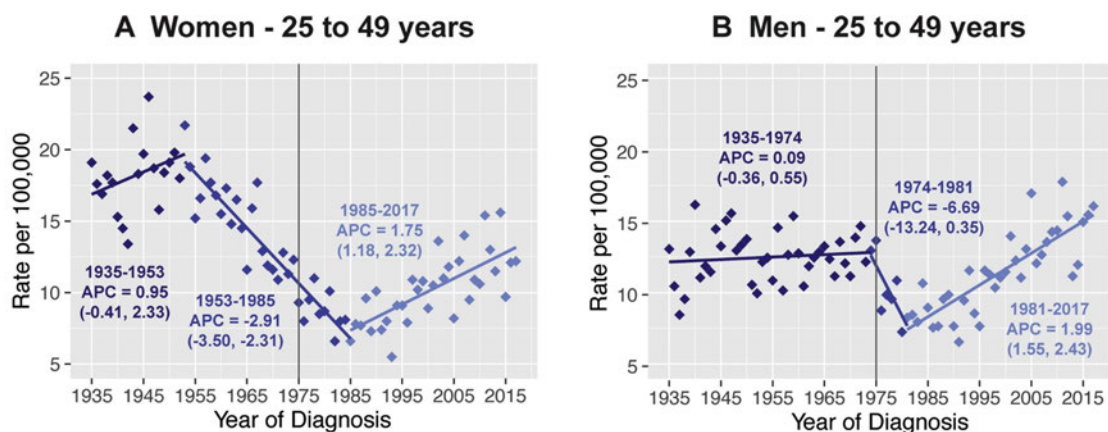


Figure 1.1: Statistics indicate an increase in colorectal cancer among people aged below 50 since 1980 in both men and women. Figure published under CC BY, adapted from [21].

As a result of its proven effectiveness and a potential shift in the recommended age for

screening [26], a high clinical demand toward colonoscopy operations is expected. However, the lack of skilled healthcare professionals, and the invasiveness and duration of the procedure (that can last more than 40 minutes) limit the use of standard endoscopy [2, 3]. Clinicians require long training to master the endoscopy technique, and the procedure's outcome highly depends on their experience. Moreover, they can suffer injuries and high levels of stress due to the excessive workload and lack of ergonomic solutions [2]. Patients often refuse to undergo colonoscopy to avoid physical and psychological stress. The application of forces to the colon wall can create loops in the intestine, stretching the mesentery and thus causing pain and patient discomfort. In younger patients, pain can be severe, especially in women, as the colon folds around the uterus and the mesentery is more likely to be stretched [27]. Sedation is required to diminish pain, but they can increase the risk of complications [28], extend hospitalisation, and raise the overall costs of the procedure [6]. Abdominal pain after colonoscopy is not rare and can be reduced by avoiding endoscope looping and minimising air insufflation. The incidence of serious adverse events like colonic perforation during colonoscopy is very low (0.1%). However, since endoscopy has assumed a more therapeutic role in the treatment of colon diseases, in complex endoluminal procedures like ESD, the rate of perforation has increased [29].

The current limitations of conventional colonoscopes necessitate the development of new robotic endoscopy platforms. These systems address a series of open challenges in endoscopy by providing more stable and comprehensive visualisation, shortening the learning curve for trainees, and improving instrument manipulation [30, 31]. In particular, robotic colonoscopy enables enhanced triangulation in many ways such as extending the workspace of the therapeutic instruments, developing collision detection algorithms [32, 33], and reducing the surgeon's hand tremor [34]. The incidence rate of looping, as one of the commonly occurring problems in colonoscopy, can be minimised if the colonoscope is being steered rather than being pushed forcibly [35].

Robotic colonoscopy platforms have already been tested in clinical trials and have exhibited promising results in terms of operating time. The mean time for performing submucosal dissection using the EndoMaster robotic system (Fig. 1.2(A)) was 18.6 minutes [36]. Using Endo-Ease (Fig. 1.2(B)) in a study, surgeons succeeded in conducting colonoscopy in 22 patients with redundant colons with a median time of 14.2 min and a high cecal intubation rate [37]. One of the advantages that robotic colonoscopy has over standard colonoscopy is obtaining improved sensory data of the current pose and state of the colonoscope. In a study conducted using ScopeGuide (Fig. 1.2(C)), it has been shown that providing a shape reconstruction of the colonoscope can remarkably reduce the operation time for cecal intubation [38]. Having an awareness of the states of a robotic system

and the ability to control them are vital for enhancing the performance of the system. In particular, sensing the states of flexible colonoscopes, such as shape estimation and distal tip localisation, is of crucial importance. Accurate sensing and control can reduce the safety concerns during operations and result in a less invasive experience for patients. Shape sensing and pose control of flexible colonoscopes can help to decrease the pain and damage to surrounding tissues [5], and guarantee the stability of the system [39].



Figure 1.2: Robotic colonoscopy platforms and technologies: (A) EndoMaster [36], (B) Endo-Ease [37], (C) ScopeGuide output of the endoscope’s shape [38]. Figure panels were reprinted under CC BY or with permission from their respective references.

One of the key emerging trends in robotic colonoscopy is the advancement of continuum and soft robotic technologies. Soft colonoscopes offer safer navigation through the curvilinear anatomy of the GI tract due to their inherent compliance, high dexterity, and variable stiffness [32]. Promising locomotion strategies in soft robotics also have the potential to reduce the overall duration of colonoscopy procedures. Consequently, shorter and less painful procedures could encourage a higher proportion of eligible patients to undergo regular screenings. Combined with integrated shape-sensing techniques, soft robotic colonoscopes represent a promising technological direction for enhancing both patient comfort and surgical autonomy in future colonoscopy practice.

In the following section, the characteristics and applications of soft robotics technologies are outlined, with a particular emphasis on their medical implementations. A major challenge related to the perception of these robots is further highlighted and addressed in the subsequent section.

1.3 Soft Robotics

Soft and flexible continuum robots are being developed for a wide range of applications and play an increasingly significant role in technological advancements across various fields. Their extensive use in areas involving human–robot interaction, such as rehabilitative devices (Fig. 1.3(E)) [40], wearable exoskeletons (Fig. 1.3(B)) [41], and surgical

tools [42], is particularly noteworthy. This growing attention toward soft robots arises from their intrinsic compliance and the high dexterity they demonstrate in diverse tasks. In surgical contexts, soft robots are increasingly employed in MIS through the design and development of soft endoscopes (Fig. 1.3(C)) [43], bronchoscopes (Fig. 1.3(A)) [44], and endovascular robots [45] (Fig. 1.3(D)). Owing to their exceptional flexibility, soft robots enable safe interaction with the surrounding environment within the human body, thereby reducing patient trauma and enhancing manoeuvrability for surgeons in MIS [17]. Their high deformability allows navigation through curvilinear tracts while minimising damage to peripheral tissues. The primary materials used in fabricating soft robots are generally low-cost and facilitate the production of miniature surgical instruments. The biocompatibility of these materials further positions soft robotics as a promising candidate for biomedical applications.

However, the development of soft robots introduces several challenges, particularly concerning proprioception and controllability. Proprioception, in the context of soft robotics, is defined as the ability of a robot to perceive its own shape through embedded sensing modalities [46]. Due to their hyperelastic characteristics, soft robots exhibit nonlinear behaviour, complicating accurate performance modelling and parameter identification. Consequently, deriving state variables that describe the shape and pose of the system through analytical models remains a significant challenge.

Furthermore, the absence of precise models hinders the implementation of closed-loop control and autonomous navigation systems. Integrating conventional perception sensors used in rigid robots may alter the mechanical properties of soft structures and introduce complex integration issues. These challenges highlight the necessity of developing sensory feedback systems compatible with the compliant nature of soft robots. In addition to providing information on the shape and pose of soft robots to enhance navigation capabilities, tactile feedback is also essential to prevent excessive force exertion. Force sensing minimises the risk of damaging surrounding environments, such as vital organs in the human body. The integration of shape and force feedback in surgical applications such as GI endoscopy ensures patient comfort [38], reduces the incidence of looping and perforation [51], and shortens the learning curve for inexperienced surgeons [52].

The existing literature categorises proprioception technologies in soft robotics according to their operational principles into EM trackers, optical sensors (e.g., Fibre Bragg Grating (FBG)), inertial, resistive, capacitive, and impedance-based types [53], as shown in Fig. 1.4. Another category of proprioception is based on integrated actuation and sensing.

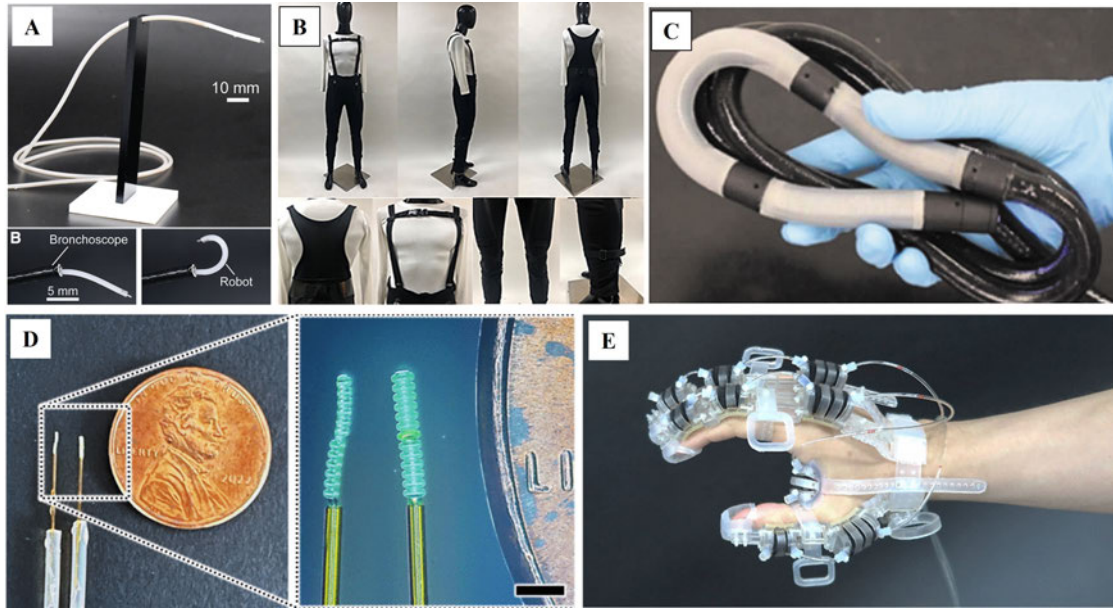


Figure 1.3: Soft robotics applications in interaction with human: (A) A soft robot for lung cancer diagnosis and therapy [44], (B) A soft lower limb Exosuit [47], (C) Morph GI soft colonoscope [48], (D) A soft robotic guidewire for endovascular interventions [49], © 2024 IEEE, (E) A pneumatically-driven soft robotic glove for rehabilitation and assistance [50], © 2017 IEEE. Figure panels were reprinted under CC BY or with permission from their respective references.

In this approach, the actuation medium, such as fluid pressure or cable tension, is used not only to drive the robot but also to infer its shape or positional information without additional modalities. Proprioceptive techniques can be integrated with model-based approaches, in which mathematical formulations are developed to describe the mechanical behaviour and kinematics of soft structures. These approaches require precise identification of the material and structural properties, which may vary over time [39]. Such models often depend on assumptions such as the **Piecewise Constant Curvature (PCC)** hypothesis or prior knowledge of external forces in Cosserat-based formulations [5]. In addition to the **PCC** assumption, where flexible bodies are represented as a series of tangent arcs with constant curvature [54], model-based frameworks such as Bézier curve representations and Frenet–Serret formulations have been extensively investigated.

Alternatively, shape-sensing frameworks can be employed to train data-driven models using **ML** and **NN**. The accuracy of these approaches relies heavily on the quality and quantity of well-labelled datasets acquired from sensing modalities [59]. As sensor outputs are often affected by noise, selecting an **NN** model that is robust to such disturbances is crucial. Furthermore, since most sensing technologies are sensitive to environmental factors such as ambient temperature, friction, and vibration, the collected datasets must exhibit strong generalisation capability. Existing sensing-based proprioceptive technologies still

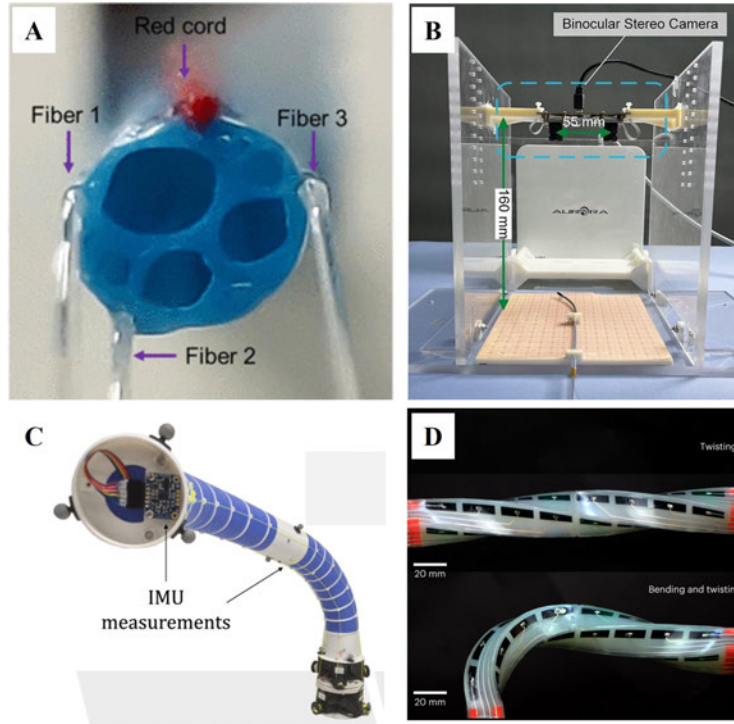


Figure 1.4: Current shape sensing technologies applicable in soft surgical instruments: (A) Optical fibres [55], © 2024 IEEE, (B) EM method [56], (C) An IMU integrated at the tip of a flexible continuum arm [57], © 2024 IEEE, (D) A capacitive flexible sensor [58]. Figure panels were reprinted under CC BY or with permission from their respective references.

face significant challenges, including susceptibility to electromagnetic interference (EM trackers), limited flexibility and high operational costs (FBG), as well as excessive wiring and hysteresis effects [60, 39].

In clinical practice, intraoperative imaging modalities such as fluoroscopy, endoscopic imaging, and ultrasound have also been explored for vision-based shape sensing [61]. These techniques provide real-time feedback on the configuration of flexible instruments but often expose patients and clinicians to high radiation doses, particularly in fluoroscopy-based systems [62, 60]. Moreover, they require bulky, costly hardware and computationally intensive post-processing procedures [38, 39]. Hybrid methods have therefore emerged, combining two or more strategies, such as model-based with intraoperative imaging [63], or data-driven with intraoperative imaging [64].

1.4 Literature Review

In this section, recent developments in soft robotics proprioception approaches, including model-based and data-driven methods, are reviewed in detail. Proprioceptive techniques

applicable to twist detection are also discussed. Many of these sensing modalities can be extended to perceive external stimuli and thereby function as exteroceptive systems, which will be outlined further into the section. The focus of this literature review is on sensing modalities that can be embedded and integrated within the mechanical structure of soft continuum instruments. Consequently, image-based reconstruction technologies are not examined here, as they fall beyond the scope of the project, despite their close relevance to the shape sensing of surgical instruments.

1.4.1 Optical

Recent research on shape sensing in flexible endoscopes has predominantly centred on optical sensing, with the most notable studies discussed here. **FBG** are widely employed for needle shape estimation [65] owing to their structural robustness and biocompatibility. An **FBG** sensor consists of a light source emitting multiple wavelengths and an interrogator that detects variations in the wavelength of reflected light. Mechanical strain, external forces, and temperature fluctuations can induce wavelength shifts, which the interrogator correlates with changes in the shape of the optical fibre and, consequently, the flexible structure in which it is embedded [39]. However, the suitability of **FBG** for medical applications requiring a large range of motion has been frequently questioned [66]. Furthermore, the high cost of optical signal interpretation and the dependence of accuracy on the number and configuration of optical cores remain major challenges.

Roodsari et al. [67] proposed a single-core optical fibre incorporating a Nitinol wire as the substrate for use as a shape sensor in endoscopic laser osteotomy. The study suggests that multi-core fibres with conventional polyamide substrates result in a larger sensor diameter and, consequently, reduced flexibility. To overcome this limitation, the authors investigated smaller single-core fibres employing Frenet–Serret equations and a kinematic model. The Frenet–Serret framework defines the tangent, normal, and binormal vectors that collectively describe a **3D** curve [68]. The results, however, indicate that further refinement is required to enhance the shape-sensing accuracy of the proposed approach.

The distance between the interrogator and optical cores in **FBG** sensors represents a limiting factor in their application to long and highly flexible endoscopes. This distance is directly correlated with the **signal-to-noise ratio (SNR)** and may adversely affect the accuracy of shape-sensing algorithms. To mitigate this issue, Lu et al. [69] developed a signal-noise reduction method based on an **Extended Kalman Filter (EKF)**, incorporating a moving-average algorithm to suppress bounded fluctuations in output signals. The approach was implemented on an Olympus colonoscope equipped with a seven-core **FBG**, as illustrated in Fig. 1.5.

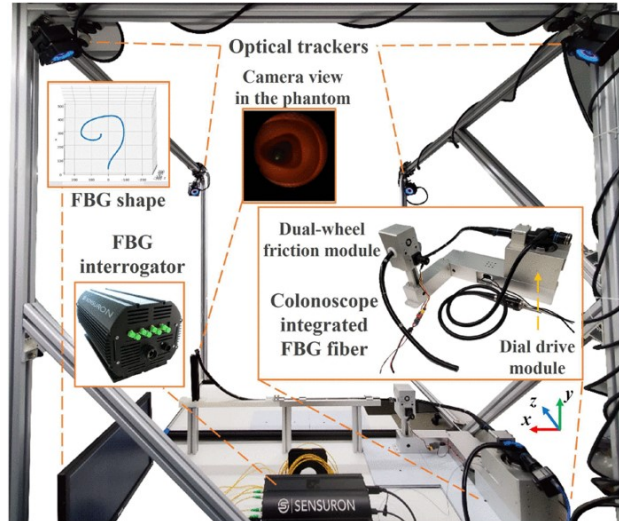


Figure 1.5: Experimental setup involving Olympus colonoscope for a multi-core **FBG** shape sensing evaluation [69]. Figure was reprinted with permission from the reference © 2021 IEEE.

Khan et al. [70] employed a multi-core **FBG** as shape sensors for flexible medical instruments such as endoscopes. Four optical fibres were integrated to reconstruct the instrument's shape by measuring the strain at each **FBG** location. Based on the wavelength of the reflected light, the curvature and torsion (twist) of the device were determined using the Frenet–Serret equations to compute the **3D** curve of the neutral axis. The four fibres were symmetrically arranged around this neutral axis, and the study assumed that temperature variations affecting the Bragg wavelength were uniform across all fibres due to their proximity. Curvature and torsion values were derived from the calculated strains at each sensor point and the relative angular orientation of one fibre to the curvature direction vector. It is important to note that these curvature and torsion values describe the fibres, not the catheter in which they are embedded. The catheter's shape was reconstructed using the Frenet–Serret formulation, which defines three orthogonal vectors, namely tangent, normal, and binormal at each point. As the distance between the fibres and the catheter's central line was negligible, the authors concluded that the fibre shape corresponded to the catheter's central axis. Thus, the curvature and torsion obtained for the fibres were applied directly to the catheter. The accuracy of the proposed approach was assessed by inserting the catheter into a **3D**-printed reference configuration and evaluating the results offline.

Liu et al. [55] proposed a fibre-optic-based shape sensing technique suitable for integration into miniaturised endoscopes such as bronchoscopes. Using distributed strain measurements, the approach captured large deformations and adopted a model-based framework grounded in the Frenet–Serret formulation to analytically characterise the curvature

and torsion of the catheter's neutral axis. The model was subsequently calibrated and validated within both **Two Dimensional (2D)** and **3D** synthetic phantom environments.

Intuitively, as the catheter length increases, cumulative errors arise. In [71], a two-segment cable-driven manipulator was employed to control and reconstruct its shape using **FBG** feedback within a model-based framework. The model comprised a sequence of rigid links connected by torsional springs, allowing simultaneous estimation of tip force via **FBG** measurements. Initially, the Cosserat model was introduced, involving the solution of nonlinear differential equations subject to boundary conditions defined by external loads and forces on the flexible structure. In this framework, the manipulator backbone was discretised into rigid links, each representing a constant-curvature arc of fixed length, bending direction, and angle. The system's stiffness was captured using rotational springs at each link junction, with stiffness determined by equating strain energy to the elastic energy stored during bending. The bending angles were computed by balancing the applied torques with the product of spring stiffness and angular displacement. The **FBG** measured strain to determine local arc curvature and tangent vectors, following a Frenet–Serret approach (see Fig. 1.6(A)). Integration of these parameters along the backbone enabled full shape reconstruction.

Hou et al. [72] employed a helically wrapped **FBG** to measure the axial elongation and bending of a soft manipulator. To extract system kinematics accurately, the study proposed an improved version of the **PCC** method. The strain measured by the **FBG** was decoupled into axial and bending components using a geometric formulation derived from the helical wrapping configuration. The enhanced **PCC** approach accounted for non-uniform stiffness distribution along the soft actuator, which can induce section-dependent variations in bending angles. The method recognises that bending at the proximal end of a section is not uniform with the remainder of the actuator, resulting in an angular offset that introduces modelling inaccuracies. This offset, illustrated in Fig. 1.6(C), cannot be determined analytically; hence, visual estimation is used to approximate its value based on actuator curvature. The offset magnitude varies according to the bending angle of the robot.

A customisable fibre optic sensor was developed in [73] and integrated into a fibre-reinforced bending actuator. The sensor facilitates the detection of twist, estimation of tip position, collision detection, and evaluation of the relative stiffness of the surrounding environment. It operates by measuring changes in the frequency of the reflected light beam. Although the sensor is not stretchable, it minimally affects the mechanical properties of the embedded robot. The sensor readings were employed to derive the tangential, normal, and binormal vectors at the tip using the singular value decomposition method

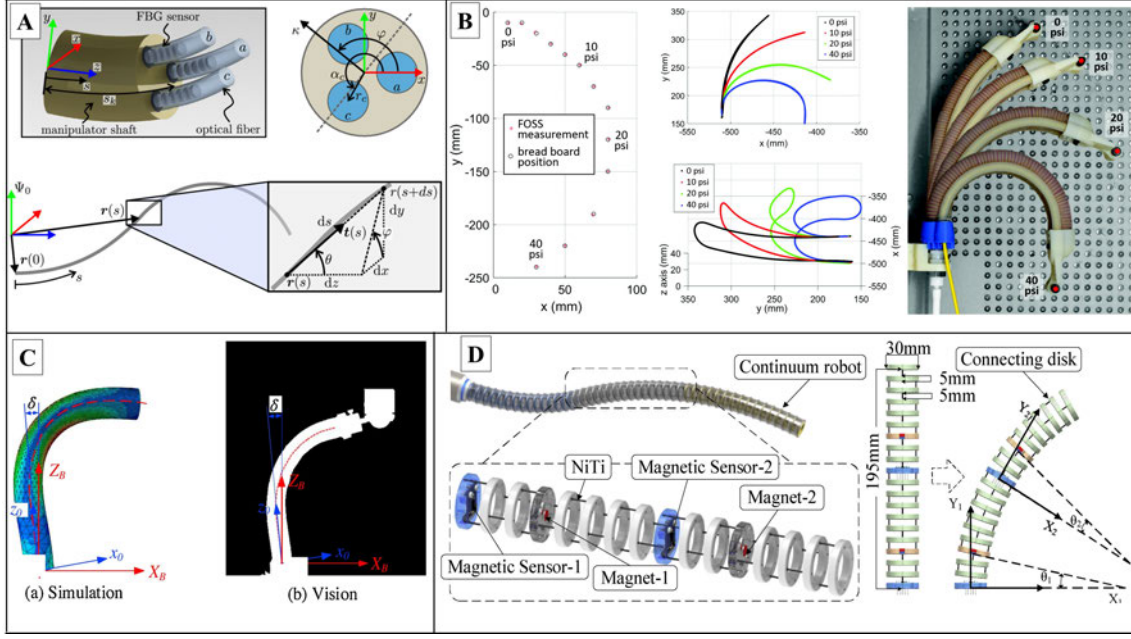


Figure 1.6: Model-based shape sensing of flexible and continuum robots: (A) Using a three-core **FBG** to measure strains along the backbone and enable shape reconstruction by identifying the tangent vector at each point [71], © 2016 IEEE, (B) Reconstructing the shape of a flexible robot in **2D** and **3D** space at different pressures using a novel optical sensor [73], (C) Demonstration of the non-uniform stiffness distribution in simulation and vision leading to inaccurate kinematic models [72], (D) The positioning of the magnet and the magnetic sensor along with the specifications of the tendon-driven robot [74]. Figure panels were reprinted under CC BY or with permission from their respective references.

[75]. These vectors were subsequently used to reconstruct the tip position and overall shape of the soft actuator, as illustrated in Fig. 1.6(B). The sensor performance was validated in **2D**, with ground truth tip positions obtained from an optical table supporting the experimental setup.

In [76], **FBG** sensors were employed to enable proprioception in a hybrid soft actuator integrating hydraulic and cable-driven mechanisms, designed for laser ablation procedures. A Koopman operator-based control algorithm was developed to regulate the tip trajectory using the **FBG** feedback signal. Another example of combining fluid-driven soft robots with **FBG** sensors is highlighted in [77]. The study introduces a soft pneumatic manipulator embedded with optical waveguide sensors to facilitate real-time tip tracking.

Optical sensors have also been investigated for the detection and measurement of twist in continuum robots. Modes et al. [78] proposed a method to determine twist, elongation, and pure bending in robots using a longitudinally embedded **FBG**. The approach was validated through simulations employing Cosserat models with a four-core fibre of circular cross-section. The authors highlight that simultaneous measurement of twist is advanta-

geous for closed-loop control, and that helically winding the fibres around the flexible arm enables twist detection in soft robots. The key contribution of the study is an analytical formulation for measuring twist and elongation of a longitudinally embedded **FBG**. The method is grounded in differential geometry and assumes that the curve is thrice differentiable. It resembles the Frenet-Serret method but is applied within the Bishop frame, where elongation can also be considered when the curvature is zero. To obtain longitudinal strain, a fibre is positioned along the neutral axis. The superposition of strains arising from elongation, bending, and twisting is employed for reconstruction, ensuring that the individual strain components do not interfere; for example, twisting does not induce fibre elongation. Cosserat rod theory is utilised to validate the method, where the shape of the curve is defined by applied torques and forces. By knowing these values, the curve and the total strains at each point can be determined. Strains at the **FBG** locations are input to the proposed model to reconstruct the shape, which is subsequently compared with the Cosserat-derived curve to evaluate error.

In the realm of data-driven shape reconstruction, Sefati et al. [79] estimated the tip position and reconstructed the shape of a surgical tool used in the treatment of pelvic osteolysis in 2D. Three **ML** algorithms were employed to predict the tip position. To reconstruct the entire shape from the tip to the base, a rigid-link model comprising multiple links connected by revolute joints was formulated. The authors compared three data-driven approaches: linear regression, a **Deep Neural Network (DNN)**, and a **Temporal Convolutional Network (TCN)**. Traditionally, shape reconstruction involves integrating slopes and positions of discrete points from base to tip, which propagates error. In this study, **FBG** sensor outputs were used as inputs to the three supervised models, with tip positions obtained from a visual system serving as targets. Following training, the predicted tip position was utilised as input for an optimisation algorithm to reconstruct the full shape. A model-based approach was also proposed, though not experimentally tested. Ground truth data were collected using two stereo cameras and colour segmentation. Hyperparameters of the **DNN** and **TCN** were optimised using an Adam optimiser. The accuracy of tip position estimation was evaluated by computing the Euclidean distance between predicted and ground truth values, while shape reconstruction error was quantified as the mean distance between predicted and actual points. Results indicated that the **DNN** performed best for tip position estimation, whereas the **TCN** achieved superior accuracy in overall shape reconstruction. The network architectures and the surgical tool utilised in this study are illustrated in Fig. 1.7(A).

Lun et al. [80] proposed a data-driven approach for reconstructing the surface of a flexible substrate using a single-core **FBG** sensor. **Finite element analysis (FEA)** was employed

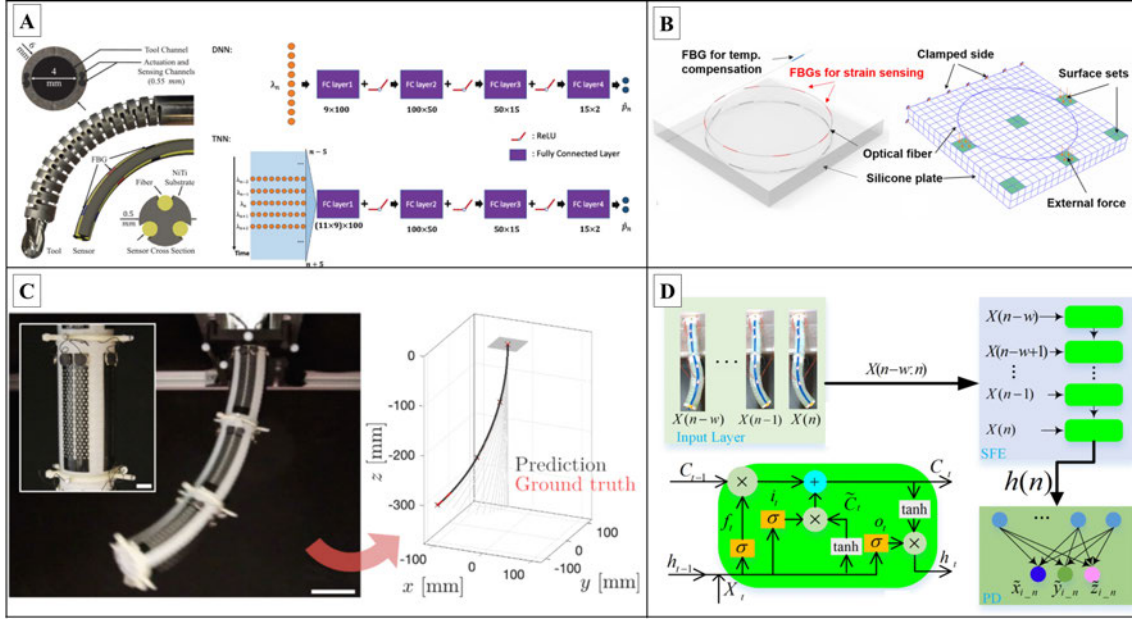


Figure 1.7: Data-driven approaches for shape reconstruction: (A) Structures of ML models used for measuring the deformation of a surgical tool for pelvic osteolysis [79], © 2021 IEEE, (B) FBG in circle forms at the top and bottom of a sensor used for surface reconstruction [80], © 2019 IEEE, (C) Data collection using a three-sectioned manipulator through piezoresistive sensors [81], (D) The structure of the network employed for shape reconstruction of a two-segmented robot under external loads by predicting the position of control points along the robot [82], © 2022 IEEE. Figure panels were reprinted under CC BY or with permission from their respective references.

to optimise sensor parameters, such as thickness, and to integrate the fibre in a circular pattern on both sides of an elastomeric rectangular substrate (see Fig. 1.7(B)). The elastomer substrate was clamped at one end, allowing manual deformation during experiments. FEA was further used to demonstrate the uniqueness of the mapping between node displacements across the surface and corresponding strain values. The novelty of this method lies in measuring strains at distributed points over the surface, rather than along the fibre as in previous studies. Temperature effects on sensor readings were compensated by an additional FBG not subjected to mechanical strain. For the data-driven reconstruction, FEA results were used to generate simulated datasets under various external loads, which were then employed to train an Artificial Neural Network (ANN). The ANN, trained using Levenberg–Marquardt backpropagation, mapped FBG measurements to the actual shape, providing reliable reconstruction. This approach offers advantages over conventional methods, including eliminating the need for calibration and compensating for fabrication defects or variations in sensor placement.

Lu et al. [83] developed a hybrid shape–force sensing system employing a helically wrapped single-core FBG for flexible ureteroscopy applications. Sparse strain signals

from the FBG were mapped to both shape and force data using three data-driven models: a Fully Connected Neural Network (FCNN), a Long Short-Term Memory (LSTM) encoder, and a One Dimensional (1D) Convolutional Neural Network (CNN). This framework enabled simultaneous estimation of curvature and axial twist, providing enhanced proprioception for flexible instruments.

In a subsequent study, Lu et al. [84] presented a robust online shape-sensing method for a multi-section extensible soft manipulator and a colonoscope using a four-core FBG. A learning-based approach incorporating a Jacobian model accounted for unknown, time-varying disturbances. Radial Basis Function Neural Networks (RBF-NNs) were employed to estimate system parameters and disturbance dynamics. This methodology enabled closed-loop shape control with guaranteed stability, based on Lyapunov theory, and convergence of estimation error to zero, without requiring prior data collection.

To overcome limitations such as self-twisting observed in FBG-based sensors [85], Jang et al. [86] introduced a novel fabrication method for colonoscopic shape sensing. The self-twisting effect, caused by plastic deformation from storage-induced pre-shaping, was mitigated using a nodal architecture composed of serially linked short polyimide tubes. Each tube was fixed at one end, while the opposite end was allowed to bend under gravitational force. FBG sensors were calibrated against predefined curvature profiles for accurate shape extraction. Additionally, a length estimation algorithm was proposed to determine the insertion depth of the endoscope. This algorithm compared live FBG signals with baseline measurements obtained from straight sections of the colon. By combining the nodal architecture and the length estimation algorithm, the method effectively reduced shape reconstruction errors caused by self-twisting and enabled reliable tracking of the endoscope's configuration during insertion.

Novel approaches combining material science or sensor-fusion techniques with optical sensors have been investigated as well. To et al. [87] developed a novel optical sensor that measures the power loss of light transmitted from an LED. The sensor combines a non-stretchable reflective material (gold) with a highly stretchable, optically transparent material, enabling deformation sensing while maintaining flexibility (see Fig. 1.8(B)). Ha et al. [88] proposed a sensor-fusion strategy combining Electromagnetic Tracking (EMT) with FBG to exploit the advantages of both modalities. While implemented on a catheter, the method is adaptable to other surgical instruments. In this approach, FBG sensors were used to reconstruct curvature, while EMT captured tip displacement. By reducing reliance on EMT, FBG sensors compensated for cumulative errors over long distances, whereas EMT corrected limitations of FBG measurements. The final shape reconstruction was achieved through Bézier curve fitting, providing accurate real-time

tracking of the catheter.

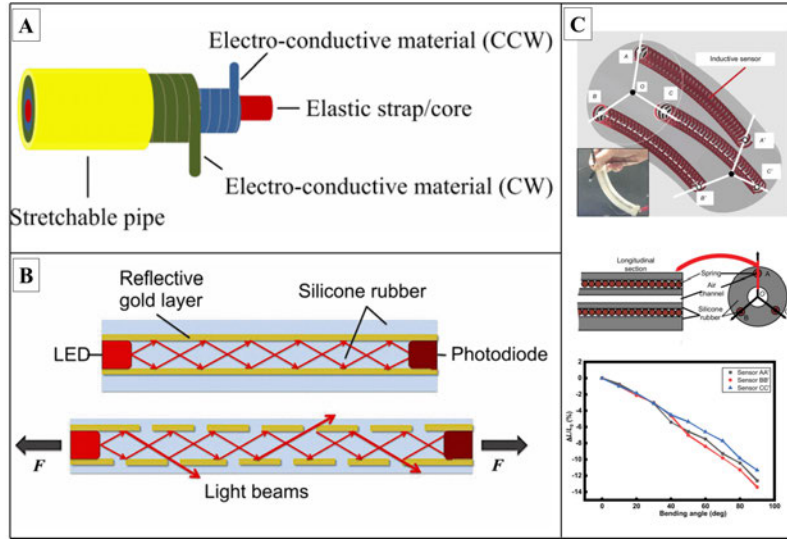


Figure 1.8: Stretchable sensors developed for shape reconstruction: (A) Internal and External components of a conductive yarn [89], © 2015 IEEE, (B) A sensor based on the combination of a stretchable (Silicone) and a non-stretchable (gold) material [87], © 2015 IEEE, (C) Using inductive fibres as fibre reinforcement and a shape sensor. The inductance is reduced upon the deformation of the robot [90], © 2020 IEEE. Figure panels were reprinted with permission from their respective references.

1.4.2 Resistive and Capacitive

Resistive sensors operate based on Pouillet’s law, detecting resistance variations to infer deformations caused by internal actuation or external loading. Common implementations include strain sensors [91] and conductive liquid-filled microchannels [92]. Despite their simplicity, these sensors often exhibit high hysteresis and limited stretchability.

So et al. [93] employed stretchable sensors composed of multi-walled carbon nanotubes to reconstruct the shape of a flexible manipulator. Several sensors were attached along the manipulator, and Frenet-Serret equations were used to derive curvature from the strain measurements, enabling full shape reconstruction.

Wurdemann et al. [89] used electro-conductive yarns composed of polyester and stainless steel to measure soft robot deflections (see Fig. 1.8(A)). The yarns exhibit a linear relationship between resistance changes and length, allowing curvature estimation under PCC assumptions. Three yarns were embedded with 120-degree offsets. A similar conductive textile sensor was developed by Cianchetti et al. [94].

Xing et al. [90] compared predicted angles obtained from yarn-based sensors with those from EM trackers. The study proposed an inductive soft sensor (Fig. 1.8(C)) as an alter-

native to fibre reinforcement in soft bending robots. These sensors, fabricated by shaping conductive fibres into spring-like structures, exhibit low hysteresis and measure robot shape and deformation.

Tapia et al. [95] demonstrated the use of stretchable resistive sensors to capture deformations in a soft pneumatic robot interacting with unknown objects. Their study emphasised optimising sensor number and placement to minimise shape reconstruction errors.

While model-based methods such as Cosserat models can predict shape based on internal and external loads, the actual forces are often unknown in practice. Consequently, data-driven approaches are preferred when shape reconstruction is performed under external disturbances. Zhao et al. [82] present a data-driven method for reconstructing the shape of a robot under both loaded and load-free conditions. The method predicts key points along the robot, which are then connected using a curve-fitting algorithm to control the robot's shape in quasi-static mode via a Jacobian-based control system. Strain gauges measure the robot's deformations, providing inputs to an NN model (Fig. 1.7(D)). Previous curve-fitting studies primarily addressed 2D conditions and neglected external loads in spatial scenarios. The method was applied to a two-segment pneumatic robot, each segment 150 mm in length, with three chambers. An RGBD camera provided labelled data to predict six control points, while additional points were used for validation. Predicted positions were fitted using a parabolic function in the XOY plane and the inverse of a sigmoid in the XOZ plane. Optimisation algorithms estimated unknown parameters within physical constraints.

In [81], a soft continuum robot comprising 12 actuators and 12 piezoresistive sensors based on a Kirigami design was developed (see Fig. 1.7(C)). The robot is fabricated from conductive silicone sheets, with sensors employed to detect its 3D shape. A combination of a kinematic model and data-driven approaches was used for this purpose. The kinematic model is based on the PCC hypothesis, while the data-driven methods involve LSTM and Multi-Layer Perceptron (MLP) networks to predict the Configuration Variables (CVs) from the sensor voltage readings. The CVs include actuator curvatures and lengths in the x and y planes, as well as segment length changes. The inverse kinematic model estimates the CVs from sensor readings, which are then used in the kinematic model to predict the robot's shape at arbitrary time steps. Curve fitting is applied to visualise the shape along multiple points of the actuator.

Loo et al. [96] proposed a multimodal sensing framework estimating both proprioceptive and exteroceptive states of a pneumatically actuated finger using flex sensors. Their approach combined Recurrent Neural Network (RNN)-based learning with an EKF and

residual adaptation scheme to improve robustness against sensor noise and drift.

Capacitive sensors measure capacitance between conductive polymers and dielectric layers, providing high-resolution morphological mapping capabilities [58]. Li et al. [97] developed a temperature-insensitive, low-cost capacitive sensor for a cable-driven continuum manipulator in orthopaedic surgery. The sensor comprises five layers, including a copper film encapsulated in polyester, distributed across multiple segments to measure curvature and tip position.

Zhao et al. [98] employed strain gauges to estimate the backbone shape of a multi-segment pneumatic soft manipulator. Their two-phase data-driven approach used an LSTM-based sequence encoder to process actuator and sensor signals, followed by an FCNN to decode three key control points per segment. Spatial curve fitting was then applied using a parabolic model in the XOY plane and an inverse sigmoid function in the XOZ plane. The estimated backbone shape facilitated a quasi-static closed-loop controller for end-effector trajectory tracking.

Conductive liquids have also been widely used to measure soft robot deformations by monitoring conductivity changes in accordance with Pouillet's law. These sensors are scalable and amenable to miniaturisation. Guo et al. [99] developed a mixture of Fe nanoparticles and gallium–indium alloy for direct printing onto PDMS and Ecoflex substrates, enabling curvature sensing. Helps et al. [100] introduced aqueous sodium chloride as a dual-purpose medium for actuation and self-sensing in hydraulically driven soft robots, using electrodes at proximal and distal ends to achieve low-cost, repeatable, and compact sensing. Similarly, Lee et al. [101] demonstrated a saline-powered foldable soft robot composed of polytetrafluoroethylene film, achieving sub-millimetre resolution through integrated self-sensing. Russo et al. [102] presented a biocompatible sensor incorporating conductive liquids, specifically NaCl. Although the liquid does not actuate the robot, it preserves the robot's mechanical properties. AC excitation was required due to NaCl's low surface tension, but the sensor remains suitable for diverse biomedical applications.

Electrical Impedance Tomography

Impedance, represented as a combination of resistance and reactance arising from alternating current excitation of a conductive medium, can be utilised to extract informative features related to system states. EIT operates by injecting electrical currents into an ionic medium and measuring the resulting voltage differentials, from which impedance distributions are estimated. By placing electrodes and defining the injection and measurement

configuration, which is referred to as the **EIT** protocol, deformations in the soft structure can be detected through local impedance variations. The adopted protocol consists of multiple **EIT** channels, each designating specific electrodes for current injection and voltage measurement.

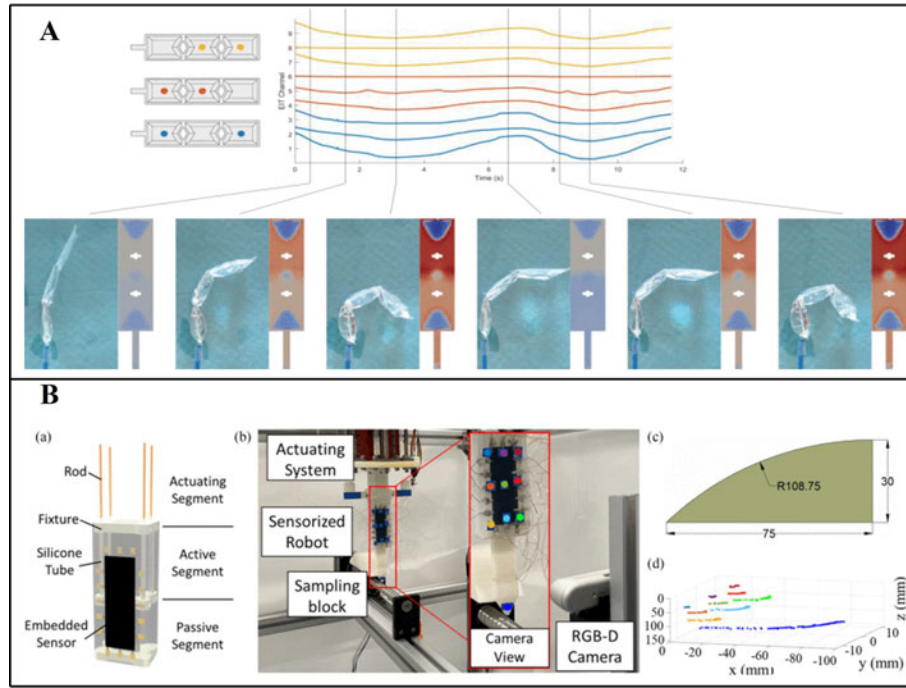


Figure 1.9: **EIT** for shape sensing in soft robots: (A) **EIT** variation across channels during the bending of a variable stiffness soft robot [8], © 2019 IEEE, (B) Surface recognition by an **EIT**-based proprioception in plane [103], © 2023 IEEE. Figure panels were reprinted with permission from their respective references.

EIT was originally developed as a medical imaging technique for reconstructing the internal conductivity distribution of biological tissues [104] in applications such as neuroimaging in stroke patients [105]. Typical **EIT** reconstructions assume a static object, which is challenging for continuum robots. There is a growing trend of extracting the information from the **EIT** data alone in robotics and artificial sensing skins. Owing to its high information density, the raw signal data prior to image reconstruction can also be exploited for sensing applications, such as in shape, pressure, and tactile sensing.

One of the earliest studies investigating proprioception in soft bending actuators using **EIT** was reported in [8], where a frequency-multiplexed **EIT** scheme was employed to assess the capability of **EIT** in detecting robotic deformation through distinct output responses. Saline was used as the conductive medium within a two-**DOF** hydraulic hinged finger and a pneumatic actuator integrated with hydraulic channels (see Fig. 1.9(A)). The method demonstrated the ability to distinctly differentiate deformations with temporal intervals of 20 ms, highlighting its potential for real-time applications. However, mechanical hys-

teresis and a dependency on electrode count for achieving higher spatial resolution were observed and identified as areas for future investigation.

The application of [EIT](#) in clinical contexts has also been explored. For example, in [\[106\]](#), a soft [EIT](#)-based balloon catheter was evaluated through numerical simulations and phantom experiments. Balloon catheters are commonly used to dilate lumens and treat obstructions. In this study, the system was utilised to detect elliptical lumen profiles during procedures such as angioplasty, where inaccurate size estimation may result in post-treatment complications. The results demonstrated that while [EIT](#) could detect occlusions with a reasonable aspect ratio, inter-channel variations adversely affected reconstruction accuracy and calibration robustness.

In [\[103\]](#), [EIT](#)-based shape sensing was integrated within an Ecoflex encapsulation using multi-walled carbon nanotubes as the conductive medium, transmitting impedance variations through 16 copper electrodes interconnected via liquid metal. A deep residual network (ResNet-18 [\[107\]](#)) was employed to map variations in the [EIT](#) sensing skin, represented as reconstructed [EIT](#) images, to electrode positions in both [2D](#) and [3D](#) (see [Fig. 1.9\(B\)](#)). Validation results demonstrated reconstruction errors below 5% for planar cases and 3% for spatial cases. However, due to the intrinsically planar structure of the sensing skin, the detection of more complex surface deformations remained limited. Furthermore, the stretchability of the sensing skin, reported at 60%, could be improved through optimisation of fabrication processes and substrate materials. The study also highlighted the susceptibility of [EIT](#) to noise, indicating the need for further measures to enhance signal quality.

[EIT](#) has additionally been investigated for tactile sensing applications. Conventional [EIT](#) approaches suffer from limited spatial resolution, requiring a higher number of electrodes to accurately reconstruct conductivity distributions and potential perturbations. In [\[108\]](#), a data-driven approach based on [NN](#) was proposed, incorporating a spatial sensitivity-aware mean squared error loss function. This method demonstrated superior spatial resolution and sensitivity compared to conventional reconstruction techniques. The approach was validated using a sensing platform comprising a conductive fabric layer and evaluated under single-touch indentation, two-point discrimination, and numerical experiments. Despite the improved resolution, the computational speed of the method was identified as a limitation for real-time deployment.

Large-area tactile sensors based on [EIT](#) were developed in [\[109\]](#) using an Ecoflex enclosure filled with soft polyurethane foam and an ionic fluid (tap water), with 16 electrodes embedded within its internal periphery. The study employed a Pyramid Scene

Parsing Network [110] to map EIT outputs to tactile information. The system was evaluated under both single- and multi-touch indentation scenarios, achieving a localisation error of 7.5 mm. However, the method exhibited location-dependent sensitivity, with reduced reconstruction accuracy in regions distant from the electrodes. This limitation was subsequently addressed in [111] through the use of a model-driven multimodal convolutional neural network that combined physics-informed voltage features with an attention-based architecture. This approach enabled simultaneous identification of contact shape and force magnitude, although the increased mathematical complexity of the mapping algorithm presents challenges for practical implementation.

Electrochemical Impedance Spectroscopy (EIS)

While EIT focuses on reconstructing the spatial distribution of conductivity through tomographic imaging based primarily on the real component of impedance, electrochemical impedance spectroscopy characterises the dynamic behaviour of electrochemical systems in the frequency domain. As impedance comprises real and imaginary components, corresponding to resistive and reactive effects, the response measured by EIS is inherently frequency dependent. By perturbing an electrochemical system with excitation signals spanning low to high frequencies, processes occurring over different characteristic time scales can be selectively stimulated and analysed, enabling the identification of distinct physical mechanisms such as charge transfer, double-layer formation, and mass transport phenomena [112].

This frequency-resolved representation is potentially beneficial for state estimation in robotics, as variations in the measured impedance spectrum may reflect changes induced by external loading or internal actuation that lead to morphological deformation of the system. To interpret these behaviours, electrochemical systems are commonly modelled using equivalent electrical circuits, which approximate complex electrochemical interactions through combinations of resistive and capacitive elements. Such models enable the decomposition of the overall impedance response according to the characteristic time constants of the circuit components, thereby facilitating physical interpretation of the measured spectra. While EIS has been extensively applied in domains such as fuel cells, corrosion science, and biosensing, its exploitation for shape and deformation sensing in robotic structures remains comparatively limited compared to studies in tissue characterisation and tactile sensing.

In [113], the electromechanical characteristics of conductive fabrics were analysed using EIS to assess their suitability as pressure sensors. To this end, the conductivity and permittivity of commercially available fabrics were characterised under varying mechanical

loading conditions by performing impedance measurements over a frequency range from 50 Hz to 1 MHz. Both numerical simulations and experimental setups were employed to capture the tensile and compressive behaviour of the fabrics.

The use of EIS for manipulation purposes was investigated in [114], where the technique was applied to identify both the type and state of liquids during manipulation (see Fig. 1.10(A)). By integrating electrodes into a gripper mounted on a robotic manipulator, the method provided a feedback signal to enhance liquid handling through analysis of the real and imaginary components of the measured impedance obtained via swept-frequency excitation. A key limitation of the approach was its restricted sampling rate, which constrained state estimation to quasi-static conditions.

Tissue scanning using bio-impedance forceps during minimally invasive surgery was proposed in [115]. The method enabled the detection of subsurface tissue inhomogeneities by employing two forceps as electrodes in a tripolar sensing configuration, as in Fig. 1.10(B). This configuration was based on the principles of the half-Schlumberger electrode arrangement, which accounts for polarisation impedance effects at the electrode–tissue interface.

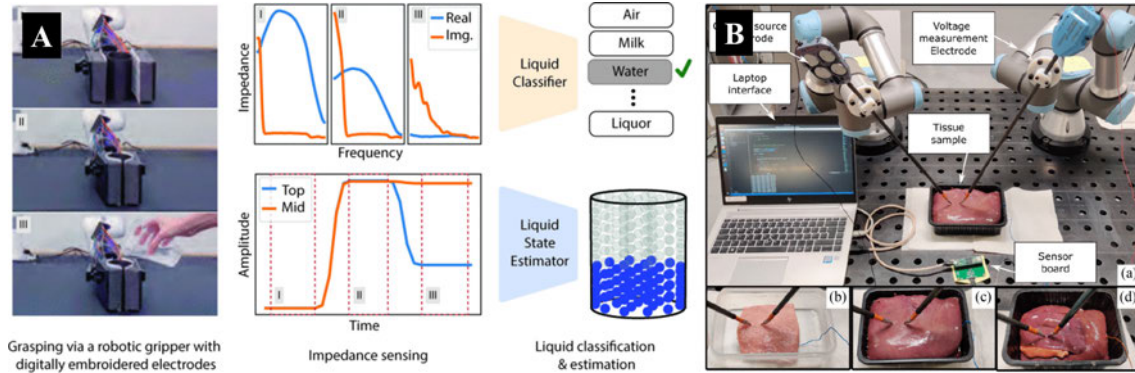


Figure 1.10: EIS application in perception and tactile sensing: (A) Robotic gripper equipped with EIS-based modality for liquid identification [114], © 2024 IEEE, (B) *Ex vivo* setup for the validation of an impedance-based tissue characterisation method [116], © 2022 IEEE. Figure panels were reprinted with permission from their respective references.

In [117], EIS was utilised to provide haptic feedback for cancerous tissue discrimination in robot-assisted surgery. Electrodes were embedded within a surgical needle designed for brachytherapy and tissue characterisation, enabling localisation of tumorous regions. Haptic feedback was generated by modelling the relationship between impedance measurements and tissue mechanical stiffness, with particular emphasis on the low-frequency impedance spectrum. The electrochemical interface was modelled using a Cole configuration comprising two resistive elements and a constant phase element. The method was

validated against manual localisation through user studies involving gelatin phantoms.

Similarly, [116] employed EIS in a tripolar configuration with a Cole-based approximation for real-time tissue identification. The proposed configuration compensated for contact impedance and was validated through finite element simulations, saline tank experiments, and *ex vivo* animal tissue studies. One of the key challenges associated with the use of EIS for tissue characterisation is the sensitivity of tissue properties to environmental factors, such as temperature, which remains an important consideration for future investigations.

1.4.3 Electromagnetic

Magnetic soft sensors [118] provide high sensitivity, straightforward integration, and immunity to visual occlusion. However, their susceptibility to external magnetic interference restricts their use in environments containing ferromagnetic materials. EMT has been widely studied for shape sensing due to its compact form factor, independence from line-of-sight, and ease of integration.

Guo et al. [74] employed two pairs of magnets and magnetic sensors to reconstruct the 2D shape of a flexible manipulator composed of connected rigid links. Each section of a specific length was instrumented with two pairs of magnets and sensors (Fig. 1.6(D)). The section was subdivided into four parts, each containing a detection system (a magnet and a sensor), except for the final part. By determining the relative bending angles of these parts from the generated magnetic fields, the positions of the start and end of each segment could be calculated. Quadratic Bézier curves were fitted to reconstruct the shape, requiring the definition of multiple control points per section. The authors further addressed the mutual interference between detection systems and investigated the optimal spacing between components to minimise field distortion.

Applications of EM trackers for twist detection have also been demonstrated [119]. In this study, three EM trackers were placed at the proximal, distal, and middle sections of the manipulator. The position and twist of the corresponding links were calculated directly from the EM outputs. A cubic Bézier curve was then fitted for shape reconstruction, requiring four control points, including the distal and proximal points. The remaining two points were estimated using an optimisation-based method, which minimised the error between the tangent lines derived from the Bézier curve and those measured by the EM trackers. The optimisation also minimised the error between the measured link lengths and the corresponding values from the curve. The Levenberg–Marquardt algorithm was employed to solve this problem. For shape reconstruction, only two of the three EM

trackers were used, while twist detection utilised all three trackers to fit a polynomial curve. A 3D model of the manipulator and the proposed approach is shown in Fig. 1.11.

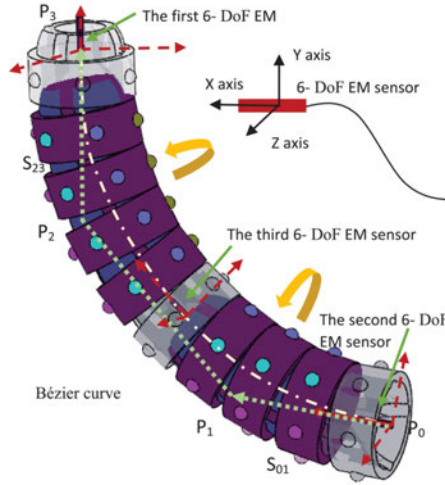


Figure 1.11: Twist and shape reconstruction through fitting a Bézier curve in a multi-sectioned manipulator. the position of EM trackers and markers for twist detection are depicted [119], © 2018 IEEE. Figure was reprinted with permission from the respective reference.

Zhang et al. [56] developed a shape sensing method for the flexible tip of a ureteroscope by placing EM sensors at the tip and base. The system adaptively selected between cubic and quartic Bézier curves, constrained by kinematic models, to reconstruct larger curvatures than previous methods [120]. Similarly, Yousefi et al. [121] combined EM sensors with Cosserat rod theory to predict both shape and external forces acting on a continuum robot in 3D space. Lv et al. [122] proposed a curve-fitting method for reconstructing the shape of flexible ureteroscopes and upper GI endoscopes using four embedded permanent magnetic sensors within the working channel. The distal shape was estimated using a non-uniform rational B-spline (NURBA) model, with parameters optimised via gradient descent under geometric constraints, including the positions of the embedded magnets and the tangent vector orientations at those points.

1.4.4 Inertial Measurement Unit (IMU)

IMUs have recently been adopted for shape sensing due to their robust pose and orientation tracking capabilities over a wide range of motion [123], outperforming other modalities that often limit the physical performance of soft robots. IMUs are inherently resistant to external disturbances such as magnetic fields, although they are prone to signal drift over time. Peng et al. [124] proposed a multi-IMU-based shape reconstruction method for a three-segment tendon-driven soft manipulator as shown in Fig. 1.12(A). The approach

was validated under static and dynamic conditions and tested with varying payloads at the distal tip. Segmental normal vectors were derived from the quaternion estimates of each IMU using an EKF, and the end-effector pose was obtained via forward kinematics under the PCC assumption.

Similarly, Stella et al. [57] introduced an IMU-based proprioceptive sensing method integrated with an analytical model based on the PCC assumption. Filtered IMU signals mitigated drift and validated the sensing approach within a closed-loop control framework. Depicted in Fig. 1.12(D), Meng et al. [125] developed a fusion-based method combining IMUs and conductive springs for shape sensing in a two-segment soft manipulator. Deformation was estimated by tracking segment length changes, providing resilience to external force disturbances. A data-driven model characterised the system, and the approach was validated in a closed-loop control scenario with a distal gripper.

1.4.5 Integrated Actuation and Sensing

Shape reconstruction can also be achieved using internal state measurements of continuum robots, such as cable tension in cable-driven mechanisms or chamber pressure in fluid-driven soft robots. Zhou et al. [128] developed a tension-deformation formulation for a cable-driven endoscope comprising rigid articulated links. Given the cable tensions, the model estimates backbone deformation by accounting for link stiffness and friction, with coefficients identified through experimental trials. Similarly, Isbister et al. [127] proposed a Cosserat rod-based kinematic model utilising tension measurements in a four-tendon-driven robot for endoscopic applications in Fig. 1.12(C). The model nonlinearly captures the effects of external and internal forces on an elastic body, accurately predicting shape under large deformations [129], while incorporating hysteresis and cable friction through a hybrid analytical–numerical formulation.

Ang et al. [126] introduced a self-sensing pose estimation method for soft pneumatic actuators relying solely on internal pressure (see the hardware setup in Fig. 1.12(B)). An ML model inferred curvature from pressure inputs, eliminating the need for external sensors and simplifying the system design.

Alternative approaches reconstruct shape from indirect measurements of key points, such as tip localisation. In capsule endoscopes, estimating the tether shape facilitates navigation and distal tip control, although these systems often lack active actuation and space for sensors. Slawinski et al. [130] proposed a shape estimation algorithm using the position and orientation of the magnetic tip, approximating the tether as a two-link robot with torsional joint stiffness. The EKF was employed to estimate link angles and stiffness

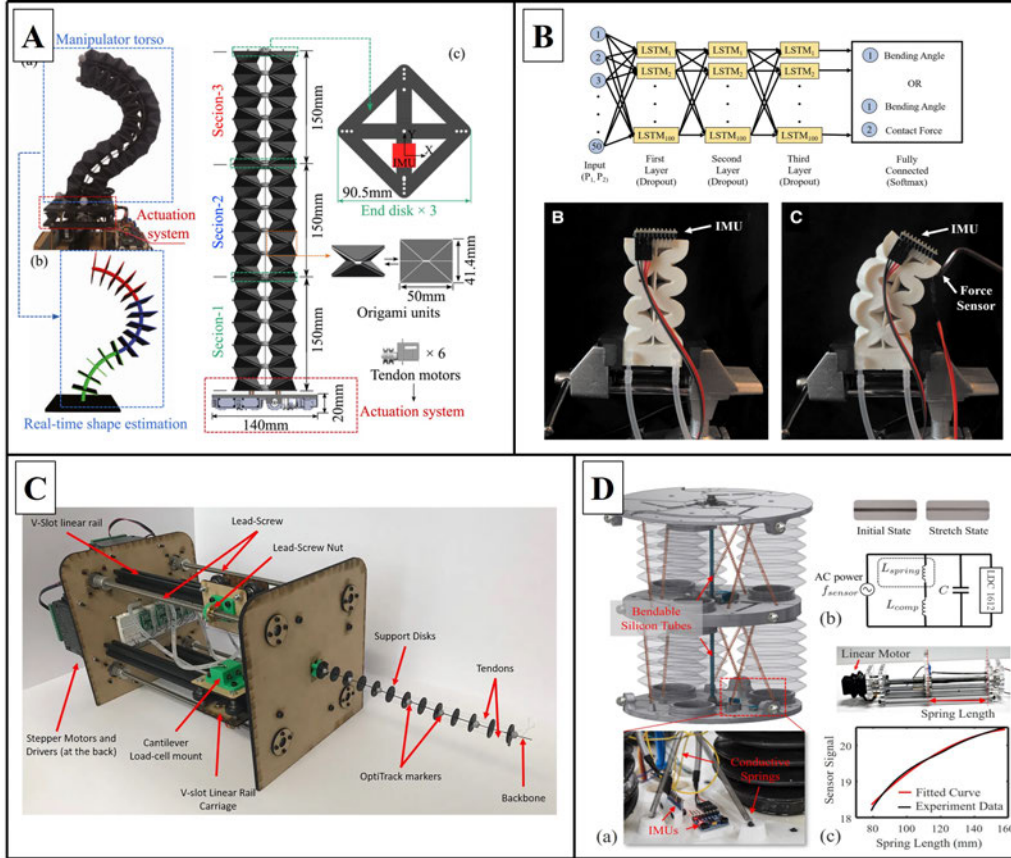


Figure 1.12: Shape reconstruction techniques using IMUs and internal states: (A) Three segmented cable-driven manipulator embedded with an IMU at the tip [124], © 2024 IEEE, (B) Utilising internal pressure and an LSTM model for self-sensing [126]. Integrated IMU at the tip samples ground truth positional data, (C) Combination of Cosserat kinematic model and internal cable tension for detecting large deflections [127], (D) Fusing IMUs and conductive springs to reconstruct the backbone shape employed as the feedback signal of a closed-loop control architecture [125], © 2024 IEEE. Figure panels were reprinted under CC BY or with permission from their respective references.

coefficients. The model assumed the PCC hypothesis. This approach can be limited when external forces acting on the tether are non-negligible.

1.4.6 Contact Force Estimation

Many of the sensing modalities reviewed for shape sensing can also be adapted to measure force or detect contact and collision. These force estimation techniques primarily rely on indirect inference of external forces from shape changes detected by embedded sensors. Selected approaches are highlighted in this section.

Qiao et al. [131] proposed a model-based framework combining Cosserat theory with an EKF to estimate the magnitude and location of external forces acting on a flexible tube

reconstructed using a four-core **FBG** sensor. The Cosserat model is particularly suited to this task, as it provides a general formulation for systems with known external forces and supports modelling of robot–environment interactions. However, direct force sensing in flexible robots remains challenging. In this study, external forces were inferred from shape deformations measured via the **FBG** strain data. Discrete curvatures at the **FBG** points were interpolated using splines to approximate the full tube shape. The **EKF** discretised the tube into a series of body frames, defining the overall configuration (position and orientation) based on concentrated loads while neglecting distributed forces and moments. The **EKF** state variables represented the magnitude and location of external forces, which were iteratively updated through linearisation and Kalman gain optimisation. This approach also enabled estimation of the minimum number of contact points using an optimisation algorithm. A key limitation lies in the need to accurately identify the robot’s stiffness, as modelling errors can affect force estimation accuracy. The experimental validation setup is illustrated in Fig. 1.13(B).

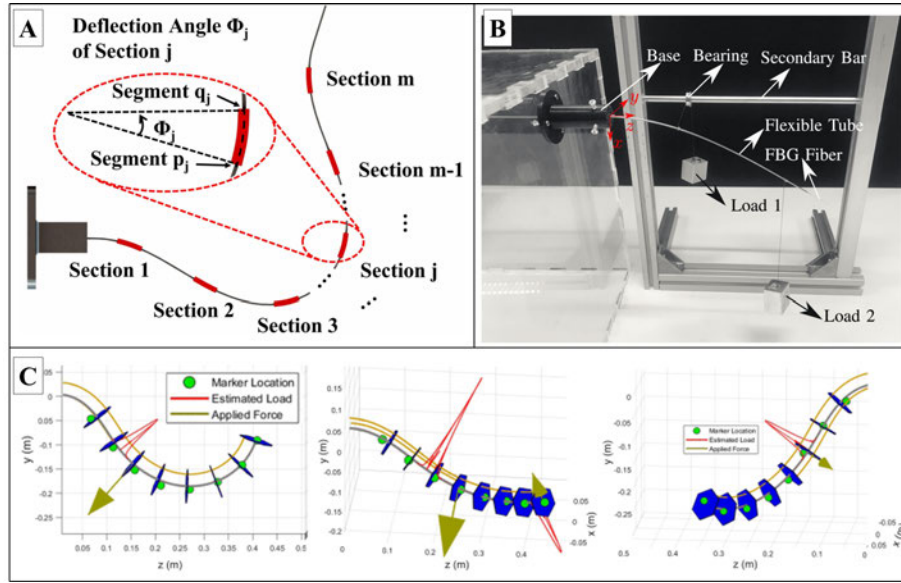


Figure 1.13: Model-based force prediction methods in the literature: (A) Estimating the tip force from shape by measuring the deflection at different sections of a slender robot [132], © 2022 IEEE, (B) Experimental setup for validation of an **EKF** based model to predict the direction and magnitude of multiple external forces [131], © 2021 IEEE, (C) Predicting the applied force (shown by yellow arrows) on a cable-driven robot in the **2D** plane and **3D** space. predictions are depicted in red spikes [133], © 2022 IEEE. Figure panels were reprinted with permission from their respective references.

Luo et al. [132] developed a method for simultaneously predicting the shape and tip interaction forces of a continuum beam undergoing large planar deflections. Their approach combined Newton–Raphson iterations with analytical Jacobian computation. As illustrated in Fig. 1.13(A), given the angular deflections along the robot and assuming

force–torque equilibrium and a geometric constraint on the total tip deflection, both the shape and interaction forces could be determined. Angular deflections were obtained using strain gauges following a calibration procedure. This method requires prior knowledge of the robot’s stiffness and assumes a passive (non-actuated) configuration.

To estimate distributed loads acting along soft continuum robots, Aloï et al. [133] proposed a model-based framework for identifying both the magnitude and location of external forces applied to slender continuum structures. Their method employs an optimisation algorithm that minimises the error between predicted and experimentally measured positions of the robot sections captured by a camera. A Gaussian parametrisation technique is introduced to represent distributed and concentrated forces along the beam, improving estimation accuracy. The approach was validated on two tendon-driven robots subjected to in-plane and out-of-plane loads. A key challenge in such inverse problems is the non-uniqueness of solutions, as multiple force–location combinations can yield similar robot shapes. To address this, the Gaussian parametrisation models external forces as a weighted summation of exponential terms representing force magnitude, position, and slope. The estimation procedure is implemented through an EKF under an unconstrained weighted least-squares optimisation, enabling efficient computation. Unlike several previous studies, the continuum robots considered here were actively actuated. The results of the proposed method are illustrated in Fig. 1.13(C).

Jin et al. [134] introduced a vision-based tactile sensing approach for endoscopy, featuring a shape memory alloy probe mounted at the distal tip of the endoscope. Two markers were placed on the probe, and changes in their relative distance under external loading were extracted from endoscopic images to estimate contact forces. Although accurate, this technique is limited to scenarios where forces act solely at the tip, excluding distributed loads along the endoscope body.

Zhang et al. [135] developed a flexible tactile sensor composed of piezoresistive elements integrated on a flexible printed circuit board, as shown in Fig. 1.14(C). Initially designed for a tethered capsule endoscope to measure contact forces on the colonic mucosa, the high compliance of the sensor makes it suitable for integration with soft continuum endoscopes for gentle tissue interaction monitoring.

A supervised data-driven framework for estimating the tip force of a concentric tube continuum robot based on its reconstructed shape was proposed by Donat et al. [136]. The method employs Deep Direct Cascade Learning (DDCL), a hierarchical neural architecture inspired by transfer learning principles. The approach enhances accuracy by integrating pre-trained networks trained on simulated datasets, making it particularly suitable

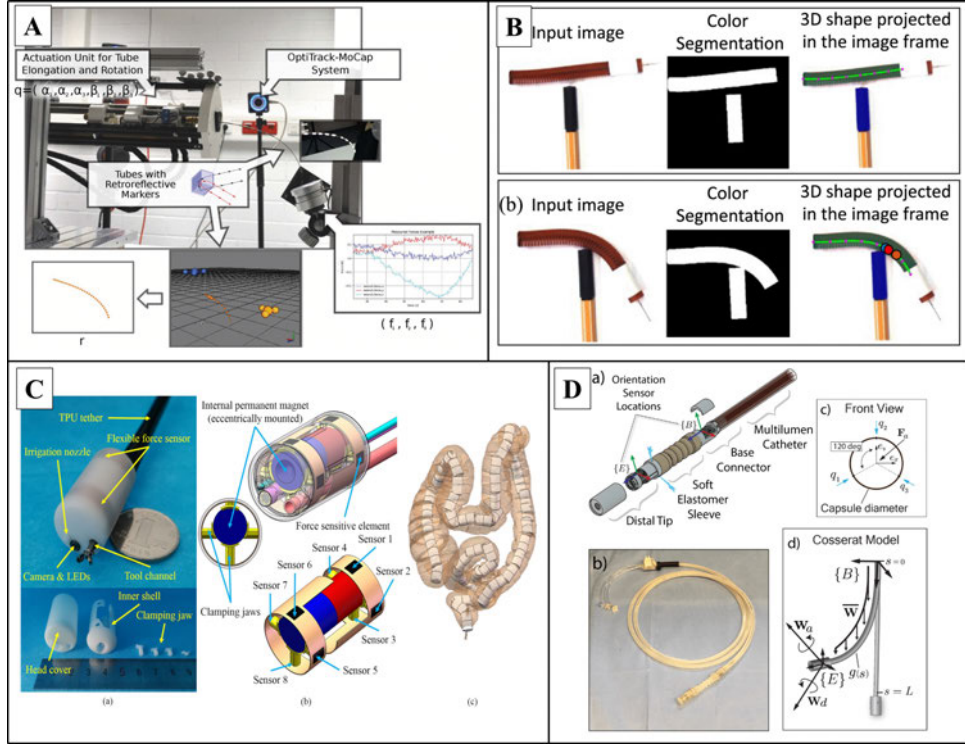


Figure 1.14: Data-driven and hybrid methods for contact force detection: (A) Experimental setup for sampling ground truth and training data for DDCL network with the aim of force prediction [136], © 2020 IEEE, (B) Collision detection and localisation of the contact point in color segmented data and the proposed data-driven approach [137], © 2022 IEEE, (C) Piezoresistive tactile sensor for a tethered capsule endoscope [135], (D) Force estimation of a flexible endoscope based on Cosserat model and two IMUs [138]. Figure panels were reprinted under CC BY or with permission from their respective references.

in scenarios where collecting extensive experimental data is impractical due to hardware constraints. The framework was validated using a multi-section constant-curvature robot and benchmarked against conventional MLP architectures with various activation functions. The results demonstrated that DDCL offers advantages in hyperparameter tuning and enables efficient layer-wise training, reducing overall training time and improving performance on small datasets. The experimental setup, incorporating an OptiTrack motion capture system and a force transducer, is illustrated in Fig. 1.14(A).

Campisano et al. [138] developed a contact detection algorithm to reduce tip position estimation error in a two-Degrees of Freedom (DOF) water-jet-actuated gastric endoscope by identifying the location—though not the magnitude—of external contact forces. The low-cost sensing unit integrates two IMUs to capture angular velocity at the tip, which is combined with a Cosserat-based kinematic model for contact detection, as shown in Fig. 1.14(D). However, reliance on velocity signals can introduce false detections due to their inherently low SNR. The method was experimentally validated in contact scenarios in-

volving rigid obstacles, with future work suggested to investigate soft-tissue interactions.

Ha et al. [137] presented a data-driven approach for collision detection and contact point localisation along the backbone of a cable-driven continuum robot. The validation setup and results are depicted in Fig. 1.14(B). The proposed method identifies discontinuities between free-motion and disturbed-motion states of the end-effector position. An unsupervised ML algorithm, AutoEncoder, was employed to distinguish these motion states, while the robot shape was reconstructed using the Frenet–Serret formulation and the k-nearest neighbour algorithm. Contact localisation is achieved by analysing the error between estimated and measured positions of embedded FBG sensors: the error increases sharply up to the contact point and remains constant beyond it. The point of contact is subsequently estimated using a piecewise linear approximation.

1.5 Discussion

In this chapter, the importance of proprioception and exteroception for continuum medical instruments, with a focus on endoscopic applications, was emphasised. Accordingly, the sensing technologies currently explored in the literature for state estimation in soft robotics were reviewed. These technologies can be classified into optical, resistive, and capacitive, IMU-based, and integrated actuation–sensing approaches. Their applications in proprioception, including twist detection, as well as in contact force estimation, were discussed. Vision-based methods such as fluoroscopy were excluded from the review, as they function as intraoperative sensing tools rather than embedded modalities.

Optical Sensing

Optical sensors, and in particular FBG sensors, represent one of the most established sensing modalities in soft robotics, with growing interest in clinical applications. These techniques infer deformation by detecting wavelength shifts induced by localised strain and temperature variations, which can be mapped to backbone shape changes of the instrument. Optical measurements may be processed using data-driven models to reconstruct instrument shape in an end-to-end manner or integrated within model-based frameworks such as Cosserat rod theory [70] and Frenet–Serret formulations [68].

Optical sensing technologies offer high strain sensitivity and inherent immunity to EM interference. Their flexibility and demonstrated biocompatibility have established them as reliable sensing solutions for minimally invasive surgical instruments in both research and commercial settings [67]. However, while the reliability of FBG sensors for bending and torsional deformation has been well demonstrated, their applicability to instruments

undergoing significant elongation remains limited due to their constrained stretchability. In addition, the high cost of optical interrogators reduces the feasibility of FBG-based sensing for disposable medical instruments. Although multi-core FBG configurations can mitigate torsional and temperature-induced effects [70], measurement accuracy degrades with increasing distance from the interrogator [69], posing further challenges for long or highly compliant instruments.

Electromagnetic Tracking

EM tracking systems capture the position and orientation of EM sensors sparsely distributed along a medical instrument, most commonly for tip localisation. Commercially available EM tracking platforms, such as Aurora, facilitate the inference of instrument configuration more directly than many alternative sensing technologies. Similar to other sensing modalities, shape reconstruction using EM tracking can be performed through both data-driven approaches [122] and model-based fitting techniques. In the literature, these model-based methods are predominantly based on Bézier curve fitting [56] and Cosserat rod theory [121].

However, EM tracking systems are highly sensitive to the presence of ferromagnetic materials in their vicinity and are susceptible to noise induced by electronic devices found in operating rooms. In addition, these systems typically offer a limited sensing volume and, due to the sparse spatial distribution of sensors, their ability to accurately reconstruct complex instrument shapes is constrained. Furthermore, EM tracking requires dedicated external hardware, which increases system complexity and cost. The additional wiring associated with these systems also hinders seamless integration into miniaturised surgical instruments.

Inertial Sensing

IMUs are compact and low-cost sensors that offer straightforward integration and accurate motion sensing, and have recently attracted increasing attention within the soft robotics community. However, these sensors are prone to signal drift and are sensitive to vibration and other dynamic effects within their operating range. As IMUs do not provide direct shape information, extracting meaningful state estimates requires complementary modelling approaches, such as PCC [57], in conjunction with sensor fusion strategies. Similar to EM tracking systems, IMU-based sensing relies on additional wiring, which limits its suitability for highly miniaturised surgical instruments.

Resistive and Capacitive

These sensing modalities reflect deformation through corresponding changes in their electrical output, typically manifested as variations in resistance or capacitance. Owing to their inherent flexibility, simple structure, and low cost, such sensors have been widely adopted in soft robotics applications. The underlying substrates of most of these sensors are stretchable materials, as exemplified by conductive textile sensors [89], liquid metal channels [92], and polymer-based sensors [93].

By measuring local strain and applying Pouillet's law, resistive sensors can provide accurate curvature estimation. However, the implementation of resistive sensing remains challenged by hysteresis, non-linear behaviour, and sensitivity to temperature variations. In addition, material ageing poses a limitation in maintaining reliable output over prolonged tension–compression cycles.

Capacitive sensors, whose operating principles are based on dielectric property variations, offer high resolution in response to morphological changes. Nevertheless, they are typically more susceptible to noise and often require additional shielding to ensure signal integrity. A common limitation shared by both resistive and capacitive sensing approaches in clinical applications is the need for complex wiring architectures, which poses significant challenges to the scalability and miniaturisation of surgical instruments.

Electrical Impedance Tomography

EIT introduces a novel sensing paradigm for the perception of soft endoscopic robots by enabling distributed sensing while allowing local adjustment of spatial resolution with reduced integration complexity. Unlike resistive and capacitive sensors, which rely on discrete strain measurements, EIT captures deformation through conductivity reconstruction based on current injection and boundary voltage measurements. These measurements can be mapped to shape variations within a flexible morphology, enabling continuous deformation sensing.

EIT supports both proprioceptive and exteroceptive sensing [108], as interactions with the environment induce shape changes that are reflected in the measured impedance distribution. This dual functionality aligns well with clinical requirements by promoting miniaturisation while preserving the structural integrity of the instrument. In addition to being low cost and immune to EM interference, EIT offers inherent biocompatibility, rendering it a safe sensing modality in the event of hardware failure or leakage.

Conventional EIT data analysis methods based on image reconstruction are limited by

low spatial resolution and the ill-posed nature of the underlying inverse problem. Recent machine learning approaches as in [103] have demonstrated improved robustness to noise and offer a means to bypass the counterintuitive inference of deformation from reconstructed EIT images [8]. Nevertheless, sensitivity to boundary conditions and variations in contact impedance can degrade the robustness of EIT-based solutions. Temperature fluctuations may also introduce baseline drift and increase measurement variability. These limitations can be mitigated through robust calibration procedures and the adoption of machine learning models with improved generalisation capability.

Compared to alternative sensing modalities (as summarised in Table 1.1), EIT offers a unique balance between structural integrity, clinical compatibility, cost-effectiveness, and signal reliability. The remainder of this thesis presents a proprioceptive sensing framework for hydraulically driven soft robots, leveraging the capabilities of EIT to facilitate clinical adoption [48]. Additional applications of EIT as a tactile sensing modality are explored through phantom studies, alongside comprehensive validation via simulation, benchtop experiments, and clinical evaluations.

Table 1.1: Summary of proprioceptive sensing technologies in soft robotics.

Technology	Advantages	Disadvantages
Optical [67, 69, 70, 55] [76, 77, 72, 73, 86]	- High spatial resolution - Immune to EM interference - Small size and biocompatible	- Cross-sensitivity - Low flexibility - High cost
Electromagnetic [74, 119, 56, 121, 122]	- Immune to occlusion - Real-time tracking - High update rate	- EM interference - Complex setup - Conditional accuracy
IMU [124, 57, 125]	- Compactness - Modular reconstruction	- Signal drift - Noise sensitivity
Integrated Actuation– Sensing [128, 127]	- Simplified design - Dual functionality	- Limited precision - Requires accurate models
Resistive, Capacitive [93, 89, 94, 97]	- Low cost - Stretchable materials	- Hysteresis & nonlinearity - Limited scalability
EIT [8, 106, 103, 108]	- Distributed sensing - Minimal wiring - Dual functionality	- Calibration variability - Configuration-dependent accuracy

Chapter 2

Planar Tip Localisation Using EIT

2.1 Introduction

As discussed in the previous chapter, shape sensing is crucial for enhancing controllability, navigation, and safety in colonoscopy. This chapter presents preliminary studies on EIT as a viable shape-sensing method for novel soft surgical robots. The objective is to evaluate the feasibility of EIT for Soft Continuum Manipulator (SCM) through proof-of-concept experiments in both simulation and hardware. These evaluations demonstrate the potential of EIT as a compliant, low-cost, low-profile, and scalable approach to shape estimation.

Building on earlier work where EIT was applied to a hydraulic hinged actuator [8], this chapter provides evidence of its practicality in hydraulically actuated SCMs. Unlike the prior study, the present work explores manipulators with higher degrees of nonlinearity, capable of continuous deformation and robotic actuation. The setup also features reduced wiring and a more compact design, making it suitable for integration as the steering portion of a flexible colonoscope.

The simulation studies employ FEA and a MATLAB interface to replicate both the mechanical deformation of the SCM and the operational principles of EIT. These simulations not only evaluate EIT responsiveness to shape changes but also examine the influence of electrode number on sensing performance. Based on the promising simulation outcomes, experimental validation was conducted by embedding EIT into an in-plane bending SCM. Initial tests manually pressurised the SCM to assess correlations between pressure, deflection, and impedance changes. A dedicated experimental platform was then developed to collect paired EIT measurements and tip positional data. An RNN was trained to establish

the mapping between these modalities, enabling tip localisation.

The contributions of this chapter are as follows:

- The demonstration of EIT in hydraulically driven SCMs.
- Development of a data-driven framework to map EIT signals to tip positions.
- Establishment of a foundation for future translation of EIT-based sensing to spatial shape reconstruction in soft robotic colonoscopy.

The remainder of this chapter is organised as follows. Section 2.2 introduces the working principles of EIT and details the simulation studies. The hardware implementation and data collection process together with the RNN model description are presented in this Section as well. Section 2.3 reports simulation and experimental outcomes, including EIT responsiveness and tip localisation accuracy. Finally, Section 2.4 discusses the implications, limitations, and future directions of this work. The contents and results of this chapter were partially adopted from [14], © 2023 IEEE.

2.2 Methods

2.2.1 EIT Working Principles

The principle of measuring the shape of the **SCM** using **EIT** relies on impedance changes caused by pressurisation with an ionic medium such as saline. Inflation of the **SCM** alters the volume of the conductive medium, leading to radial and/or axial expansion depending on the **SCM**'s design. By placing an array of electrodes within this volume, impedance variations induced by deformation can be measured through the **EIT** protocol. This behaviour is governed by Pouillet's Law:

$$R = \rho \frac{L}{A} \quad (2.1)$$

where R is the impedance, ρ the electrical resistivity, L the length of the conductive path, and A its cross-sectional area. The law indicates that resistance increases with path length and decreases with cross-sectional area. As the **SCM** deforms, the parameters L and A vary accordingly. By employing a sufficiently complex mapping model, a relationship can be established between these geometric variations and the **SCM** configuration. According to Equation 2.1, the configuration of the manipulator can thus be inferred from the impedance outputs of the **EIT** system.

In practice, the electrodes can be embedded into a **Flexible Printed Circuit (FPC)**, which conforms to the **SCM**'s geometry due to its inherent material flexibility. Further details of the **FPC** design and the impedance measurement protocol are provided later in this chapter.

2.2.2 EIT in Simulation

In the simulation implementation of **EIT**, two key computational challenges arise [139]:

1. **Forward problem:** computing the corresponding measurement voltages given the conductivity distribution of the medium and the injected current.
2. **Inverse problem:** estimating the conductivity distribution from the electrode voltage measurements.

In this work, only the forward problem is considered, since the objective is to measure the **EIT** outputs as the flexible body deflects. To solve the forward problem computationally, the **Finite Element Method (FEM)** solver EIDORS is employed. This framework solves

the quasistatic Maxwell partial differential Equations 2.2 [140].

$$\nabla \cdot (\sigma \nabla u) = 0, \quad (2.2)$$

where u is the electric potential and σ is the conductivity distribution. Maxwell's equations are subject to the complete electrode model boundary conditions 2.3.

$$u + z_k \sigma \frac{\partial u}{\partial n} = V_k \quad \text{on electrode } e_k, \quad (2.3)$$

where n is the outward normal to the boundary, z_k is the contact impedance of the k th electrode, and V_k is the corresponding electrode potential.

EIDORS [141] allows the user to define boundary conditions, including electrode configurations and model geometry, using FEM meshes. The conductivity of the medium can also be specified. The simulation workflow therefore proceeds as follows:

1. Generate different deformations of the SCM,
2. Mesh the deformed geometries,
3. Import the meshes and electrode placement data into EIDORS,
4. Compute the corresponding output voltages.

Deformations of the SCM's working chamber (saline-filled chamber) are generated using Computer-Aided Design (CAD), in combination with an equivalent rigid-link kinematic model [142] (Fig. 2.1(a)). In this model, the SCM is represented by a four-link rigid manipulator operating in a plane with three DOFs, denoted q_1 , q_2 , and q_3 , which correspond to the relative rotations at joints one, two, and three, respectively. A constant-curvature assumption is applied between consecutive joints. The forward kinematics of the rigid model, expressed in terms of link lengths and joint rotations, enables the tracking of any arbitrary point along the manipulator in 2D space. The model is given by Equation 2.4:

$$\begin{aligned} X_{EE} &= l_{r2} \sin(q_1) + l_{r3} \sin(q_1 + q_2) + l_{r4} \sin(q_1 + q_2 + q_3), \\ Y_{EE} &= l_{r1} + l_{r2} \cos(q_1) + l_{r3} \cos(q_1 + q_2) + l_{r4} \cos(q_1 + q_2 + q_3), \end{aligned} \quad (2.4)$$

where:

- X_{EE}, Y_{EE} are the Cartesian coordinates of the end effector,
- $l_{r1}, l_{r2}, l_{r3}, l_{r4}$ are the lengths of links 1–4, respectively,

The first link, L_{r1} , is fixed along the vertical y-axis, while the remaining links undergo planar rotations governed by their respective joint angles.

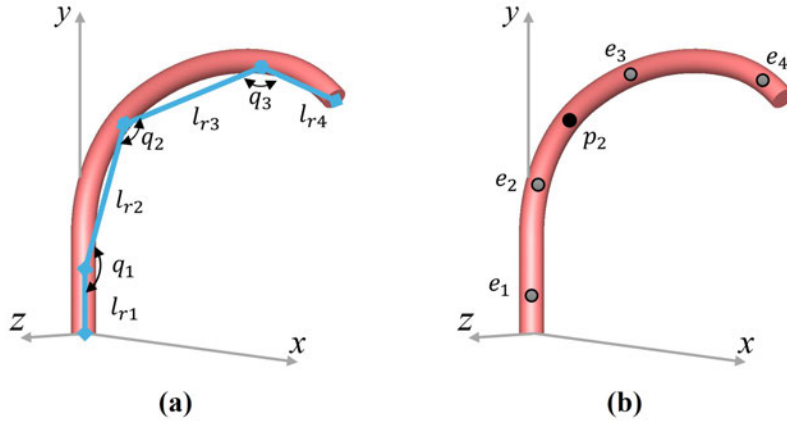


Figure 2.1: Simulating deformations: (a) Equivalent rigid kinematic model defined by variables $q_1 - q_3$, (b) Location of p_2 for EIT measurements in the case of four electrodes ($e_1 - e_4$) on the SCM.

Although numerous kinematic models based on the PCC assumption have been proposed for soft robots, this work adopts a simplified variant of the rigid-link formulation introduced in [143]. In the original framework, the soft robot structure is represented as a sequence of constant-curvature segments, each modelled by an equivalent rigid mechanism comprising two rotational joints at the ends and two prismatic joints in between, forming an RPPR chain. Accordingly, for the purposes of proof-of-concept evaluation and minimising modelling variability, the following modelling assumptions were adopted:

Assumption 1. *The soft robot was segmented into four sections, each satisfying the PCC hypothesis. The rigid counterparts of these segments were modelled using only two rotational joints, with the prismatic joints present in the original RPPR formulation omitted. This simplification reduces model complexity while preserving the dominant bending behaviour.*

Assumption 2. *Each segment was assumed to bend with a constant radius of curvature, undergoing relative rotation with respect to adjacent segments. This assumption follows directly from the PCC formulation and enables tractable kinematic modelling.*

Assumption 3. *The soft robot was assumed to be radially constrained by fibre reinforcement, with limited radial elasticity of the underlying material. Consequently, radial expansion was neglected, and the robot configurations were modelled in SolidWorks.*

This model was used to generate 50 distinct configurations in the CAD software by assigning $q_1 \in [5, 30]^\circ$, $q_2 \in [10, 50]^\circ$, and $q_3 \in [10, 35]^\circ$. The link lengths were set as $l_{r1} = l_{r4} = 12.5$ mm and $l_{r2} = l_{r3} = 37.5$ mm. The working chamber was assumed to

have a uniform cross-section along the **SCM** length with a diameter of 5 mm. The forward kinematics were also used to determine the electrode positions during **SCM** bending, which were later employed in EIDORS for mesh refinement around the electrodes.

Following the deformation simulations in SolidWorks, the configurations were exported in .STEP format along with the electrode coordinates. These files were used to generate tetrahedral meshes in Gmsh, an open-source **3D** finite element mesh generator [144]. The Python API of Gmsh was employed to control mesh sizes and perform refinements. Based on the electrode positions from the forward kinematic model, the mesh size was set to 0.2 mm within 2 mm of the electrode surface, increasing linearly to 2 mm at a distance of 7 mm. Fig. 2.2(a,b) illustrate a meshed configuration with refinements around four electrode positions visualised in Gmsh and EIDORS, respectively. The configuration corresponds to $q_1 = 10^\circ$, $q_2 = 18^\circ$, and $q_3 = 15^\circ$ with 349,175 tetrahedral elements.

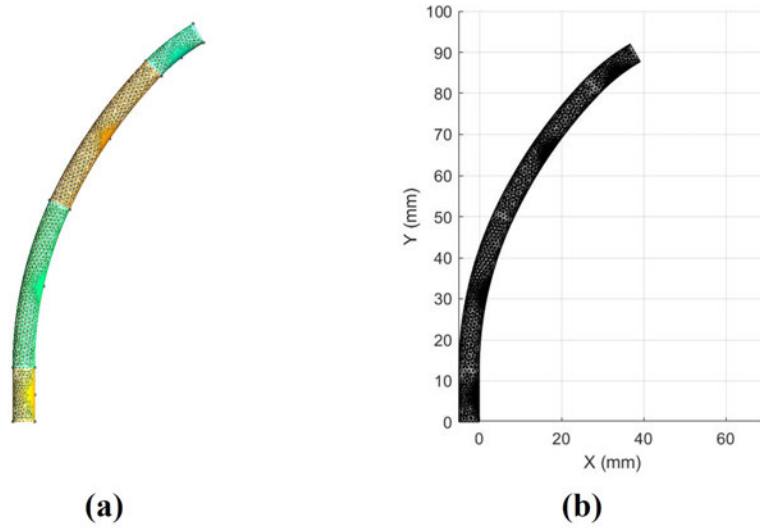


Figure 2.2: Meshed geometry with 349,175 tetrahedral elements visualised in: (a) Gmsh, (b) EIDORS.

This process was repeated for cases with seven and eleven electrodes. The electrode distribution was designed to increase the number of electrodes within each curvature with even spacing between the electrodes of each segment. The base segment was assigned a single electrode, as it is assumed to remain fixed during deformation. For the remaining segments, the number of electrodes was allocated in proportion to segment length, with segments l_{r2} and l_{r3} receiving a higher electrode density than l_{r4} due to their greater length.

Based on this strategy, in the four-electrode configuration, each segment was assigned a single electrode (as in Fig. 2.1(b)). In the seven-electrode case, the base segment retained one electrode, while two electrodes were allocated to each of the remaining segments.

In the eleven-electrode configuration, the base segment again retained a single electrode, segments l_{r2} and l_{r3} were each equipped with four electrodes, and segment l_{r4} was assigned two electrodes. The number of electrodes on l_{r4} was kept lower due to its length being approximately one third of that of l_{r2} and l_{r3} . The increase in electrode count on l_{r2} and l_{r3} from two to four was introduced to induce more pronounced variations in the EIT outputs. Configurations with a higher number of electrodes were not examined due to the increased computational cost associated with meshing and simulating the corresponding EIT models in EIDORS.

In addition to evaluating the responsiveness of EIT to deformations, this enabled investigation of the effect of electrode number on the range of output variations. The simulation protocol was defined as follows. For each excitation electrode pair, the same set of measurement pairs was used. The excitation began with electrode one as the positive pole and electrode two as the negative pole. Measurement pairs were selected as every two consecutive electrodes (e.g., 1–2, 2–3, and 3–4 for the case of four electrodes). For the subsequent excitation pair, the positive pole remained fixed at electrode one, while the negative pole shifted from electrode two to electrode 3. The measurement pairs remained unchanged. This procedure was repeated until the negative pole reached the final electrode. At that stage, the positive pole shifted to the next electrode, and the same pattern was repeated.

In the results section, EIT variations are examined by selecting the channel in which the current is injected through the first and last electrodes, with voltage measurements reported from the two electrodes immediately adjacent to point P2 (e.g., e_2 and e_3) (Fig. 2.1(b)). Point p_2 corresponds to the location of the second joint in the equivalent rigid manipulator, which was selected because it undergoes relatively large deformations near the mid-curve of the SCM. Finally, the meshed configurations and electrode positions were imported into EIDORS, where the forward problem was solved with conductivity defined as 16 mS/cm.

To evaluate the responsiveness of EIT in practice to detect deformations under experimental conditions, a hardware setup consisting of an in-plane bending SCM composed of silicone and integrated with a 13-electrode FPC, a manually operated saline-filled syringe for actuation, and a data acquisition interface for EIT measurements (impedance spectroscopy) were employed and detailed in the next sections.

2.2.3 Fabricating a Planar SCM

The **SCM** investigated in this chapter is a prototype of the soft, flexible, steerable section of a colonoscope, designed for in-plane bending with a single **DOF**. This restricted design simplifies the evaluation by limiting actuation to a single chamber, enabling a clearer assessment of **EIT** performance. In subsequent chapters, **EIT** will be extended to **SCMs** with multiple **DOFs**. The manipulator was fabricated by casting Ecoflex 00-50 silicone rubber (Smooth-On, Inc., Macungie, PA, USA) into a 3D-printed mould, following the procedure described in [145, 146]. As illustrated in Fig. 2.3: A. The mould, consisting of a lid, a half-round rod, and a main body with a double helical ridge along its length, was designed such that the extracted manipulator contained an inflatable chamber with a semi-circular cross-section. B. An inextensible polyester layer was incorporated to constrain axial elongation. C. Fibre reinforcements were added to restrict radial expansion. D. A second phase of moulding was carried out to secure the inextensible layer and fibre reinforcements in position. E. A second phase of moulding was carried out to secure the inextensible layer and fibre reinforcements in position.

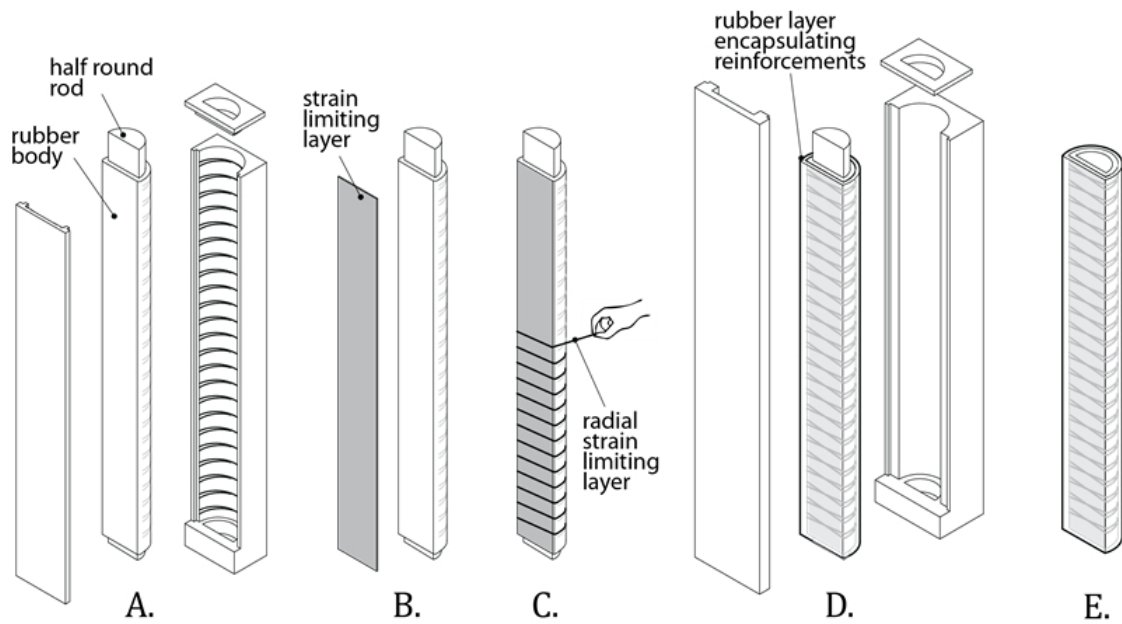


Figure 2.3: Fabrication process of the one-DOF SCM [146].

The resulting manipulator in Fig. 2.4(b) is 100 mm in length, with a wall thickness of 4 mm and an overall diameter of 12 mm, ensuring that the **SCM** remains within the size constraints of conventional colonoscopes (reported within the range of 11.1 mm to 15 mm in [6, 147]). The fibre reinforcement pitch was set at 5 mm (Fig. 2.4(a)), and the diameter of the inflatable (working) chamber was 4 mm (Fig. 2.4(c)).

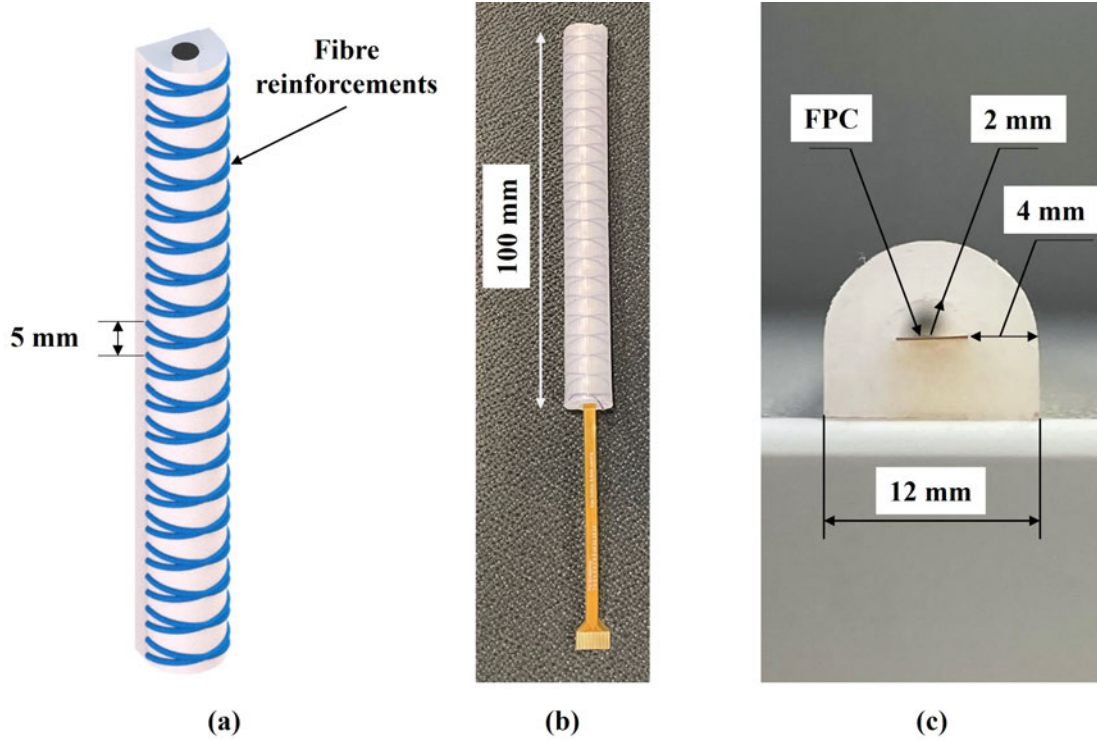


Figure 2.4: Fabricated one-DOF SCM: (a) Schematic after fibre reinforcement, (b) Prototype with integrated FPC, and (c) Cross-section with dimensional details.

2.2.4 EIT Embodiment

Prior to capping the distal end of the actuator, a flexible but non-stretchable FPC was integrated into the actuation chamber. The FPC's underlying layer is composed of a polyimide film with 13 electrodes arranged linearly with a constant spacing of 6.5 mm. The flexibility of the film ensured minimal influence on the mechanical behaviour of the actuator. To reduce contact impedance between the conductive fluid and the electrodes, the pads were coated with ENIG (Electroless Nickel Immersion Gold), thereby lowering noise in the voltage signals. The actuator was pressurised with 0.9% saline (90,000 ppm), which served as the conductive medium for EIT measurements. The FPC was designed with an effective length of 95 mm, matching the active length of the SCM, and a width of 4 mm, corresponding to the width of the working chamber (see Fig. 2.5(a)). After finalising the outline based on these specifications, the FPCs were fabricated by PCBWay (China), shown in Fig. 2.5(b).

To integrate the FPC, the polyimide film was placed on the flat side of the semi-circular cross-section of the actuator (see Fig. 2.4(b,c)). Under the assumption of pure bending, this surface corresponds to the neutral axis and is therefore suitable for accommodating a non-stretchable FPC. To secure its position and minimise noise caused by electrode

displacement, both ends of the FPC were glued using Sil-Poxy adhesive.

Impedance measurements were obtained using an impedance spectroscopy (Quadra, Eliko, Estonia). The system injected alternating currents of 5 mA sequentially using a Time-Division Multiplexing (TDM) scheme. Impedance measurements were recorded across a switching frequency range of one kHz to 349 kHz, from which a frequency of 11 kHz was selected for the subsequent experiments in this chapter based on the frequency sweep analysis provided in the results section.

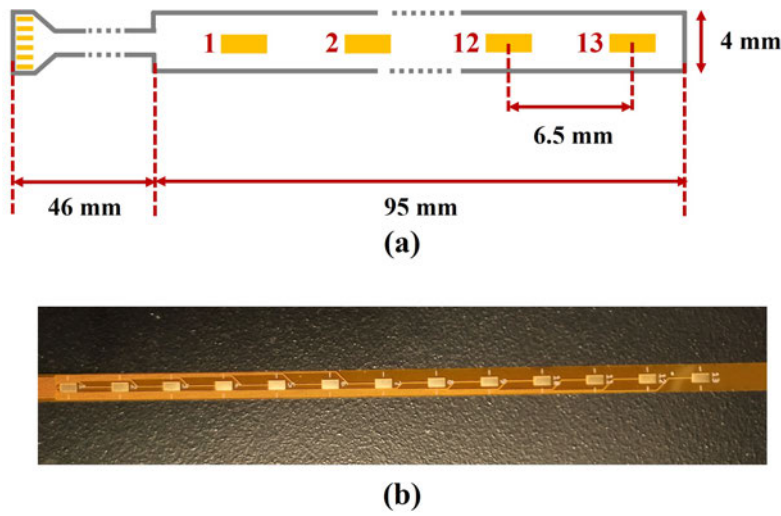


Figure 2.5: FPC for the planar SCM: (a) Schematic of the outline and corresponding dimensions, (b) Finalised FPC fabricated by the manufacturer.

2.2.5 Control System

The SCM was pressurised using a saline-filled syringe (BD Plastipak, 50 ml), connected to the proximal end via a hydraulic tube with an inner diameter of 2.4 mm and an outer diameter of 4 mm (Saint-Gobain, France). Sealing was ensured by tightening the inlet with a cable tie. To enable motorised pressurisation, a hydraulic pump system was designed. The pump system comprised a linear stage consisting of a leadscrew (4-start, 2 mm pitch) driven by a stepper motor (NEMA 23, Ooznest) with a resolution of 1.8° and a torque output of 12.6 kg·cm. The motor was controlled by a μ Stepper S-lite board (uStepper IVS, Denmark), which integrates an onboard encoder and was programmed via the Arduino IDE. Pressure was measured using an MS5803-14BA sensor (TE Connectivity, Switzerland) with a resolution of 1 mbar. Absolute pressure data were acquired throughout the experiments and transmitted to an Arduino Due (Arduino, Italy) at 20 Hz via an I2C interface. The complete control system is shown in Fig. 2.6(a).

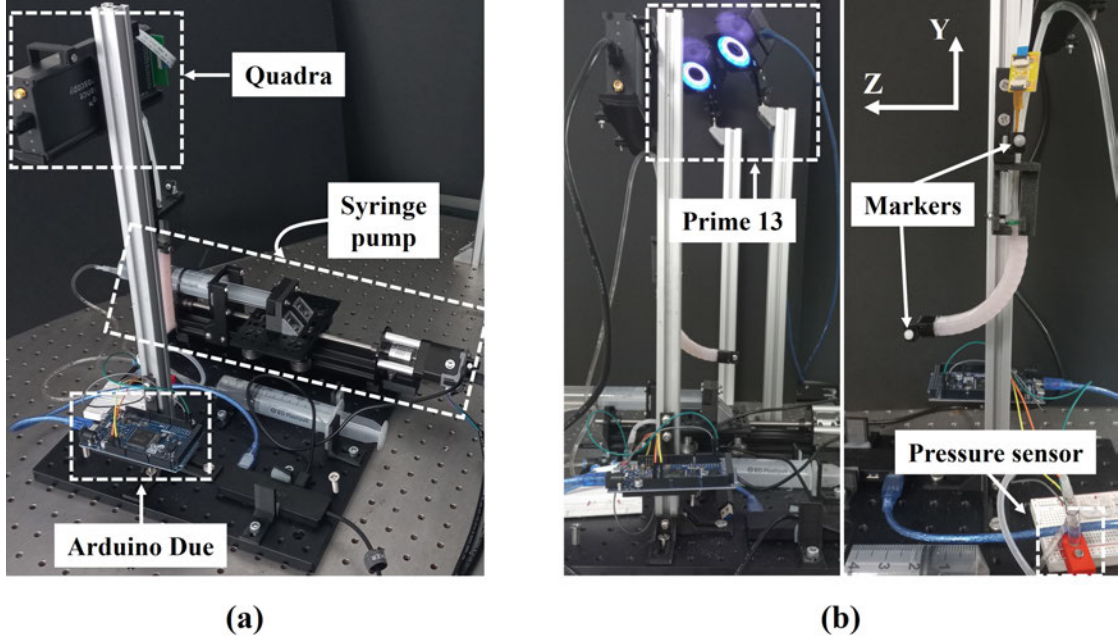


Figure 2.6: Experimental hardware to pressurise, monitor and track the planar SCM: (a) Control system, (b) Motion capture tracking system.

2.2.6 Experimental Setup

Two sets of experiments were conducted to evaluate the performance of EIT under experimental conditions. The first aimed to verify whether EIT is sensitive to deformation when integrated into the SCM, while the second assessed the robustness and repeatability of the data for potential use in data-driven frameworks.

Experiment One – EIT Responsiveness

In the preliminary tests, the syringe was operated manually to pressurise the SCM. Pressure was increased gradually and held constant at arbitrary intervals to assess system stability. At six distinct pressure levels, the manipulator's shape deformation was recorded using a camera, enabling correspondence between visual configurations and their EIT measurements. The bending angles of each configuration were subsequently computed using image processing and SolidWorks.

The adopted protocol consisted of 84 EIT channels, sampled at 0.4 s intervals. This configuration allowed for the evaluation of various current injection pairs and provided spatial coverage across the working chamber. Fig. 2.7(a) illustrates the protocol schematics. Injection electrode pairs are denoted by Ex_i , where i refers to the channel index ($i \in [1, 84]$). For example, $Ex_{1,2,3,\dots,7}$ denotes that the electrode pairs are used as current injection pairs for EIT channels one through 7. The symbols $+$ and $-$ represent the positive and negative

poles of each pair. Measurement electrodes are denoted by V_i , with distinct colour coding or line styles (e.g., dashed, dash-dotted) for clarity. Due to space limitations, only the first seven channels of the protocol are shown in Fig. 2.7(a). For subsequent channels, the measurement pairs remain unchanged while the negative pole of the injection pair is shifted to the next electrode.

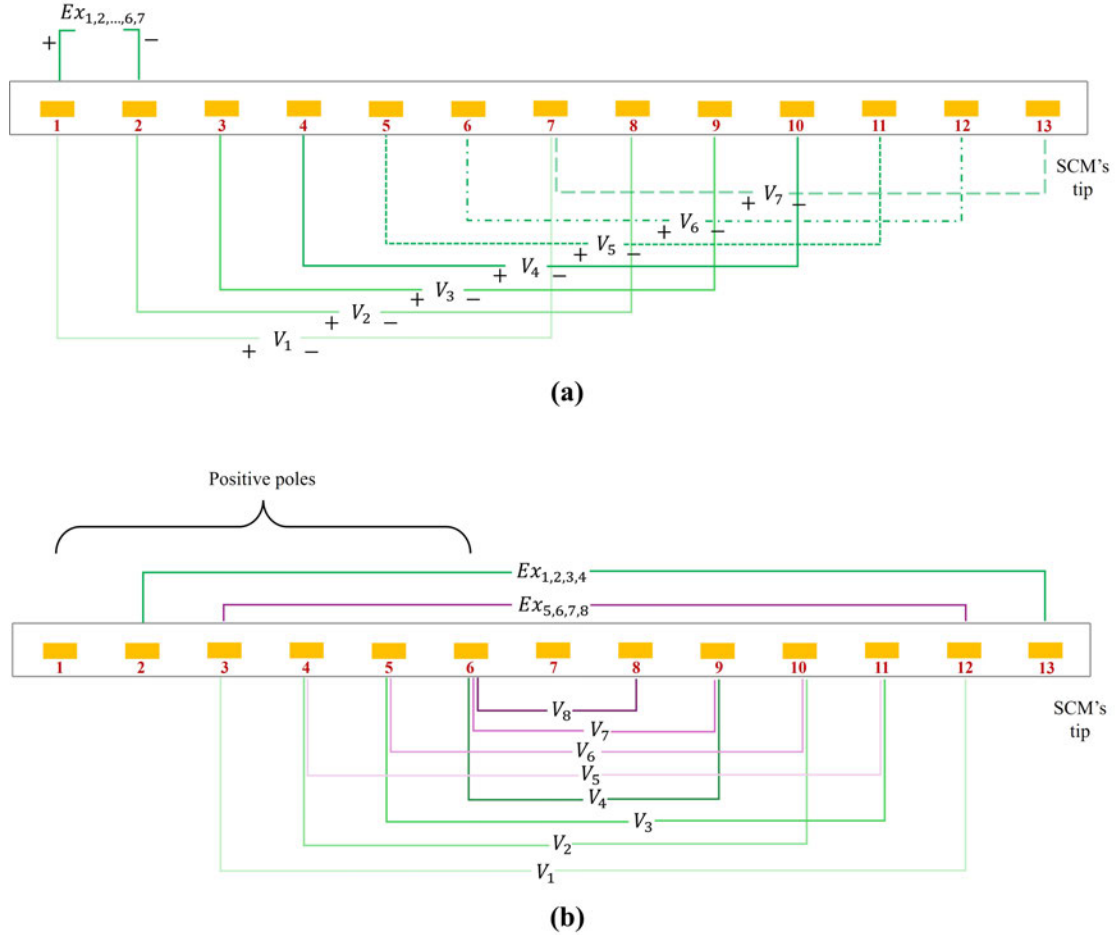


Figure 2.7: EIT protocols employed for validation experiments: (a) 84-channel protocol for EIT responsiveness tests, (b) 8-channel protocol for tip localisation data collection.

Experiment Two – Tip Localisation with EIT

In the second experiment, the planar SCM was mounted on an aluminium stand using a 3D-printed fixture, allowing motion in the vertical plane. Two reflective markers were placed: one on the rigid base and another at the distal end of the actuator, enabling optical tracking of the tip position using an OptiTrack motion capture system (Prime 13, NaturalPoint, USA) (see Fig. 2.6). To attach the distal marker, a lightweight rigid component was fixed to the SCM tip.

Pressure data were recorded at 20 Hz via serial communication with MATLAB, using

the syringe pump system to apply random triangle-wave pressure signals with amplitudes ranging from 1163 to 1664 mbar. Optical data were collected at 125 Hz and used both as training input and as ground truth for tip position predictions. The total recording time was 20 minutes, corresponding to 17,553 motion samples. The dataset was divided into training and testing subsets with a ratio of 75:25.

For this experiment, an 8-channel EIT protocol was adopted, sampled at 0.05 s intervals. Compared to the 84-channel protocol used in Experiment One, the reduced channel count decreased training time while capturing shape-change information across the SCM's length (see Fig. 2.7(b)). Electrode one was excluded due to high contact impedance caused by electrode oxidation. Measurement pairs were colour-coded according to their associated injection pairs, with all positive electrodes located on the left half of the outline (electrodes 1–6).

The 1- and 4-channel protocols were defined as subsets of the 8-channel protocol. For the 1-channel protocol, Ex_1 was used with the same V_1 measurement pair. For the 4-channel protocol, $Ex_{1,2}$ corresponded to $Ex_{1,2,3,4}$, and $Ex_{3,4}$ to $Ex_{5,6,7,8}$, with measurement pairs V_1 , V_6 , V_5 , and V_7 , respectively.

2.2.7 Machine Learning Model

To predict the tip position of the SCM, a learning framework capable of handling nonlinear, hyperelastic, and time-dependent behaviour is required, as the silicone rubber used in the SCM renders its dynamics time variant [148]. RNNs are well-suited for modelling sequential data [149]. However, RNNs suffer from limitations such as vanishing and exploding gradients due to their restricted memory capacity. To overcome these issues, LSTM networks were introduced [150]. By incorporating additional gating mechanisms, LSTMs improve memory retention and capture long-term dependencies in sequential datasets [151]. In this study, the training and predictions are carried out using an LSTM-based framework inspired by [152]. Further mathematical details and background can be found in [149, 153].

The LSTM model is trained using EIT impedance values and pressure measurements as the feature space, while the output (target space) corresponds to the 2D coordinates of the tip relative to the base marker. The network architecture comprises five layers: an input layer of size nine (one pressure input and eight voltage inputs for the eight-channel protocol), a dropout layer with rate 0.1 to reduce overfitting, an LSTM layer with 50 hidden units, a fully connected layer with two outputs (tip y - z coordinates), and a regression output layer (see Fig. 2.8).

Training is performed for a maximum of 2000 epochs with mini-batches of size 512. The initial learning rate is set to 0.005 and follows a piecewise schedule, decreasing by a factor of 0.1 every 125 epochs. Prior to training, all data are normalised using the mean and **Standard Deviation (SD)** of the training set. The Adam optimiser [154] is employed for weight updates. The network is implemented in MATLAB 2022a.

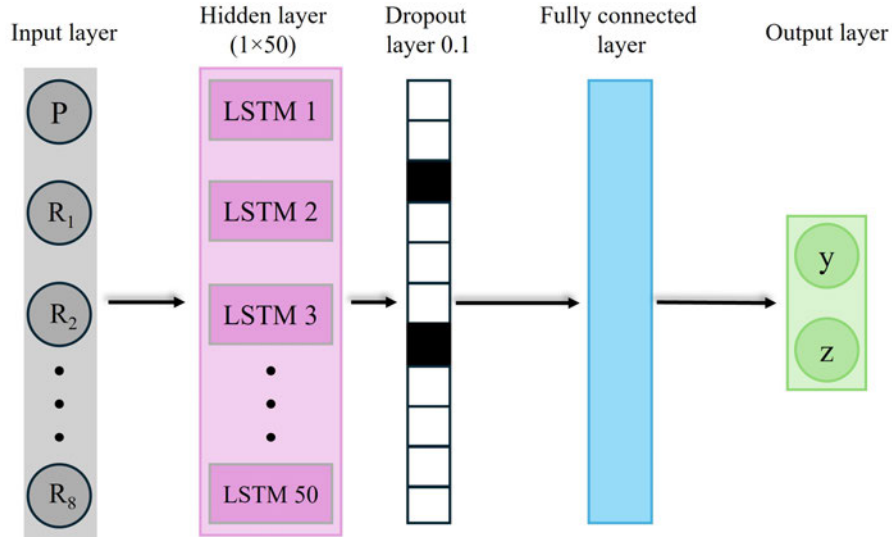


Figure 2.8: **LSTM** architecture trained with pressure and 8-channel **EIT** data to reconstruct the **SCM**'s tip position in **2D**.

2.3 Results

2.3.1 EIT in Simulations

According to the adopted protocol, the total number of channels for each electrode configuration is as follows: four electrodes: 18 channels; seven electrodes: 126 channels; eleven electrodes: 550 channels.

To assess the most affected portion of the **SCM**, the case with eleven electrodes and the middle configuration ($q_1 = 17.5^\circ$, $q_2 = 30^\circ$, $q_3 = 22.5^\circ$) was considered. The selected channels correspond to those where the first and last electrodes were used as the injection pair (1–11). The associated channel numbers range from 91 to 100 (out of 550 in total), as labelled in Fig. 2.9. Channel 91 corresponds to the measurement pair 1–2, while channel 100 corresponds to 10–11. Among these, the middle channel (95) produced the highest voltage of 60.37 mV, followed by 43.11 mV and 41.62 mV in channels 92 and 98, respectively. These values represent the absolute **EIT** outputs without calibration to the initial configuration.

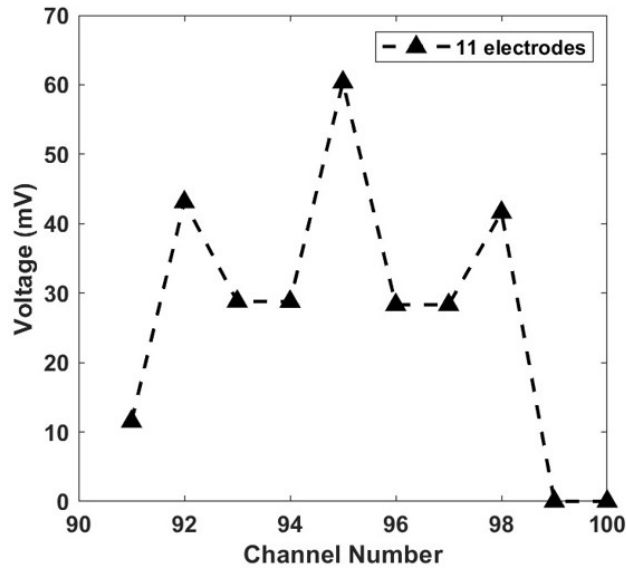


Figure 2.9: EIT outputs across the SCM for the middle configuration in simulation in the 11-electrode case.

From Fig. 2.9 and the electrode arrangement in Fig. 2.1, it can be observed that the lowest outputs occur at the proximal and distal ends of the SCM. The outputs of channels 93, 94, 96, and 97 are less significant, as their measurement electrodes lie on the same link. By contrast, larger variations are observed between electrodes positioned on consecutive but separate links, as in channels 92, 95, and 98. Similar behaviour was observed in the cases with four and seven electrodes, where the middle channel consistently delivered the highest voltage.

Given this, the output of the middle channel was adopted to investigate the influence of electrode number on the output range. As the number of electrodes increases, the spacing between them decreases, thereby affecting EIT outputs. The corresponding middle-channel indices are as follows: four electrodes: channel 8; seven electrodes: channel 33; eleven electrodes: channel 95. In Fig. 2.10, the x -axis represents the 50 configurations simulated by EIDORS and SolidWorks. Configuration one corresponds to $q_1 = 5^\circ$, $q_2 = 10^\circ$, $q_3 = 10^\circ$, whereas configuration 50 corresponds to $q_1 = 30^\circ$, $q_2 = 50^\circ$, $q_3 = 35^\circ$. The figure demonstrates that, regardless of the electrode count, the output voltage increases as the SCM deforms (progression to higher configurations).

Since the SolidWorks simulations did not model SCM expansion or cross-sectional changes, the voltage increase is mainly attributed to the growing distance between measurement pairs, driven particularly by changes in q_2 . As the distance increases, resistance rises according to Pouillet's law, and consequently the voltage increases due to Ohm's law.

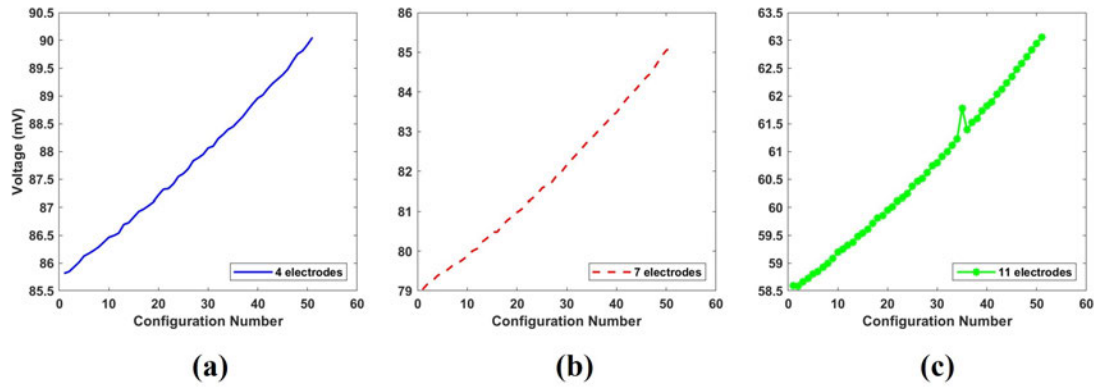


Figure 2.10: EIT measurements for different numbers of electrodes between configuration number one and 50: (a) Four, (b) Seven, (c) Eleven electrodes.

It can further be observed that both minimum and maximum output voltages decrease as the electrode number increases. Specifically, the minimum voltage recorded at configuration one was 85.81, 79.03, and 58.58 mV for four, seven, and eleven electrodes, respectively. Similarly, the maximum voltage recorded at configuration 50 decreased from 90.05 to 60.06 mV as the electrode number increased from four to eleven. This reduction is explained by the smaller spacing between consecutive electrodes at higher electrode counts, encompassing a lower volume of conductive medium.

In all three cases, the voltage increased approximately linearly, which likely reflects the linear increments in joint angles across configurations. The average ratios, expressed as ΔV per configuration, were 0.085, 0.122, and 0.089 mV per configuration for four, seven, and eleven electrodes, respectively. These results suggest no direct correlation between electrode number and output voltage ratio, although an optimal number may exist. Finally, the sudden jump of 0.55 mV at configuration 35 in Fig. 2.10(c) can likely be attributed to the mesh model and refinement effects.

2.3.2 EIT Responsiveness in Practice

Frequency sweep analysis of the EIT output from Experiment 1 was conducted to characterise the signal response of a representative channel (Channel 25) across 15 excitation frequencies and multiple dataset instances (see Fig. 2.11). The results indicate that an excitation frequency of 11 kHz lies within the mid-frequency plateau region, where the impedance response demonstrates clear separation between distance steps while remaining relatively insensitive to high-frequency capacitive attenuation effects.

Frequencies above approximately 80 kHz exhibit a pronounced amplitude roll-off and reduced inter-step contrast, whereas lower frequencies show increased variability and re-

duced stability across trials. Consequently, 11 kHz provides a suitable trade-off between sensitivity, measurement stability, and signal-to-noise ratio, supporting its selection for subsequent analysis. The impedance changes measured by EIT in the first experiment,

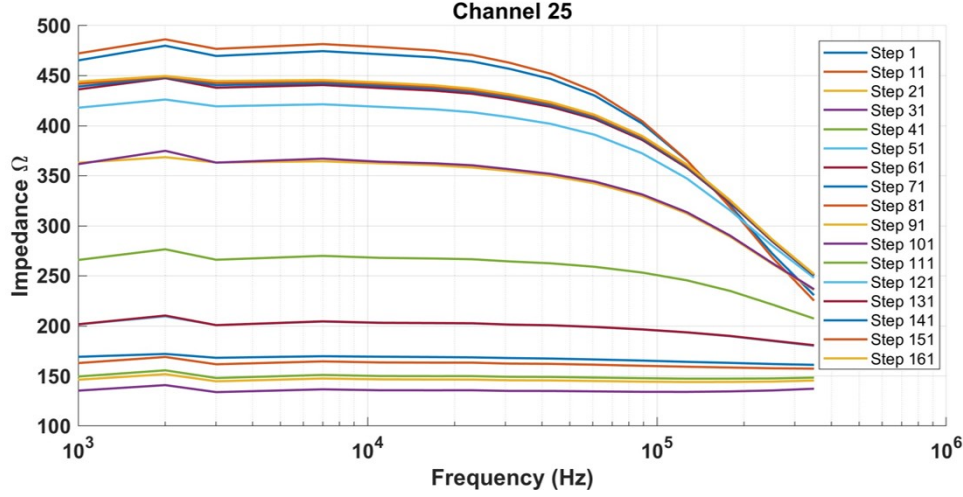


Figure 2.11: Frequency sweep analysis of EIT output from Experiment one.

corresponding to channel one, are plotted against six configurations (Fig. 2.12(a)) in which the applied hydraulic pressure was maintained constant (see Fig. 2.12(b)). The EIT values were normalised with respect to the maximum impedance of channel one (603 Ω) in Fig. 2.13(a), obtained in the first configuration of the SCM with a bending angle of 3.7°.

The results demonstrate that the EIT system detects deformations and shape variations. Fig. 2.12(b) illustrates the normalised EIT output at a frequency of 61 kHz for channel one, corresponding to the excitation and measurement electrodes indicated in Fig. 2.13(b). As the bending angle increases from 3.7° in configuration *a* to 76.1° in configuration *f*, the impedance decreases from 603 Ω to 334 Ω , equivalent to a 45% reduction. The most significant drop (16%) occurs between configurations *a* (603 Ω) and *b* (508 Ω), whereas the smallest change (5%) is observed between configurations *e* (354 Ω) and *f* (334 Ω). While the bending angle intervals between consecutive configurations are not uniform, the impedance decreases appear to occur at a diminishing rate as the bending angle increases. This observation may suggest that the EIT output ratio decreases at larger bending angles.

For comparison, the EIT output of channel 25 is also plotted (Fig. 2.13(a)), with the corresponding electrode locations shown in Fig. 2.13(b). In both channel one and channel 25, the impedance consistently decreases as the actuator bends. This reduction can be attributed to the inflation of the inner chamber during pressurisation, which increases its

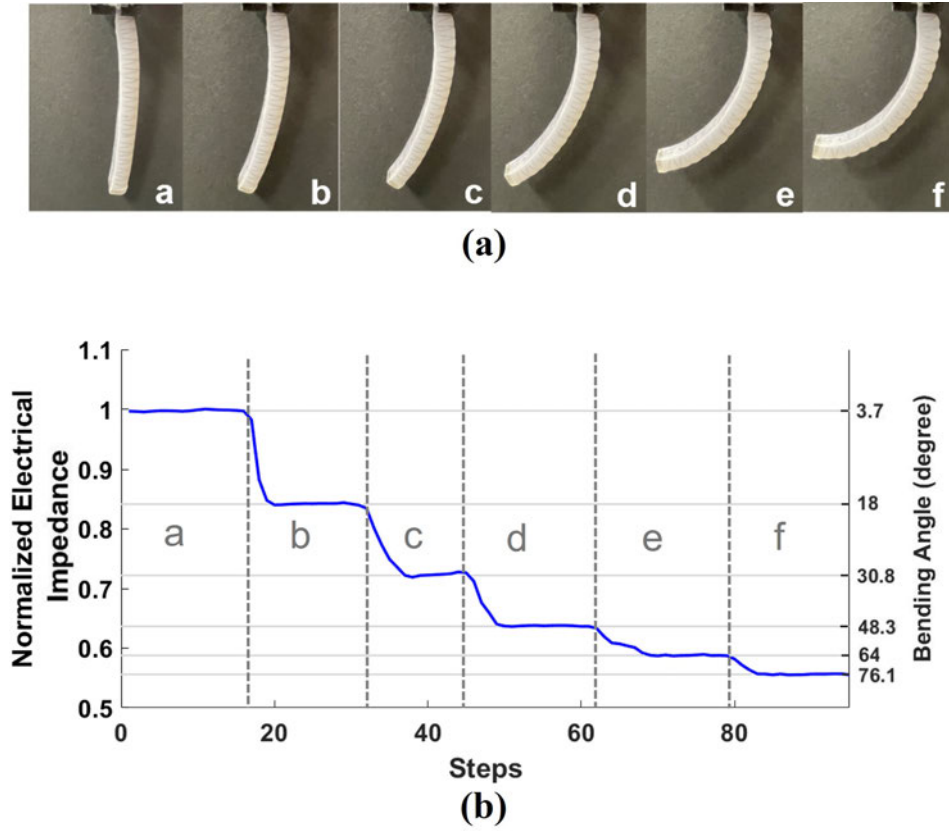


Figure 2.12: EIT evaluation on detecting deformations: (a) Six distinctive configurations, (b) Corresponding channel one EIT output normalised with respect to its maximum value.

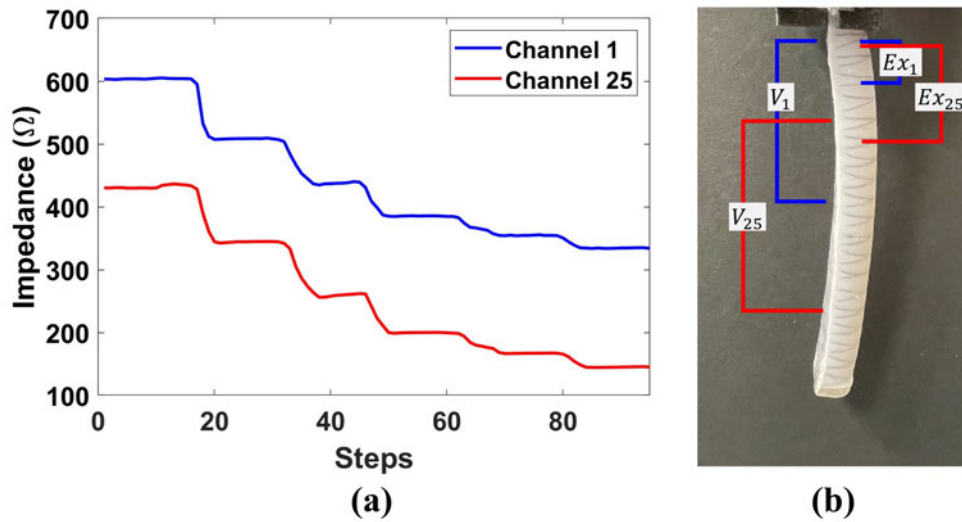


Figure 2.13: Comparison between different EIT channels' output: (a) Impedance values in channel one and channel 25, (b) Corresponding excitation and measurement portions with respect to the SCM.

cross-sectional area. Unlike in simulation, where the distance between electrodes varies, the **FPC** used here is non-stretchable, keeping the electrode spacing constant. Consequently, the increase in cross-sectional area leads to a decrease in impedance according to Pouillet's law.

Although both channels show similar trends, the magnitude of impedance in channel 25 is lower than in channel 1. This can be explained by two factors. First, the overlap between the excitation and measurement regions is smaller in channel 25 than in channel one, resulting in reduced current density through the measured region. Second, as also observed in simulation results, deformation tends to be more pronounced in the middle section of the **SCM** than at its proximal or distal ends. Since channel 25 corresponds to the middle section, greater expansion in this region results in a lower impedance.

2.3.3 EIT for Tip Localisation

EIT-Pressure relationship

In the previous experiments, the relationship between the **EIT** output and the configuration of the **SCM** was evaluated. However, in this section, for the first time, the correlation between the **EIT** output and the input pressure is investigated. A pressure signal with random amplitude, shown in Fig. 2.14(a), was applied to the **SCM** to collect the training and testing datasets for tip localisation. To observe the response of the **EIT** to this signal, Fig. 2.14(b) illustrates both signals plotted together using a dual-y-axis representation for better visualisation. The **EIT** signal shown corresponds to the third channel of the eight-channel protocol in Fig. 2.7(b), recorded at a switching frequency of 11 kHz. It can be observed that as the pressure increases, the **EIT** output decreases, indicating an inverse relationship between the two. For instance, at step 100, the pressure decreases from 1597 mbar to 1165 mbar at step 193, while the **EIT** increases from 754.16 Ω to 1853.87 Ω over the same period.

An increase in pressure implies a larger bending angle of the **SCM** due to inflation. In the previous experiment, it was observed that as the bending angle increases, the **EIT** value decreases. Therefore, the behaviour of the **EIT** signal under random pressure excitation is consistent with earlier observations, reinforcing the validity of the relationship. In Fig. 2.14(b), the variations in **EIT** are proportional to the changes in pressure. For example, between steps 100 and 193 (green region), a pressure drop of 432 mbar results in a 1099.71 Ω increase in **EIT**, whereas between steps 250 and 300 (grey region), a 199 mbar drop (46% of the previous decrease) leads to a 677.31 Ω increase in **EIT** (62% of the previous increase).

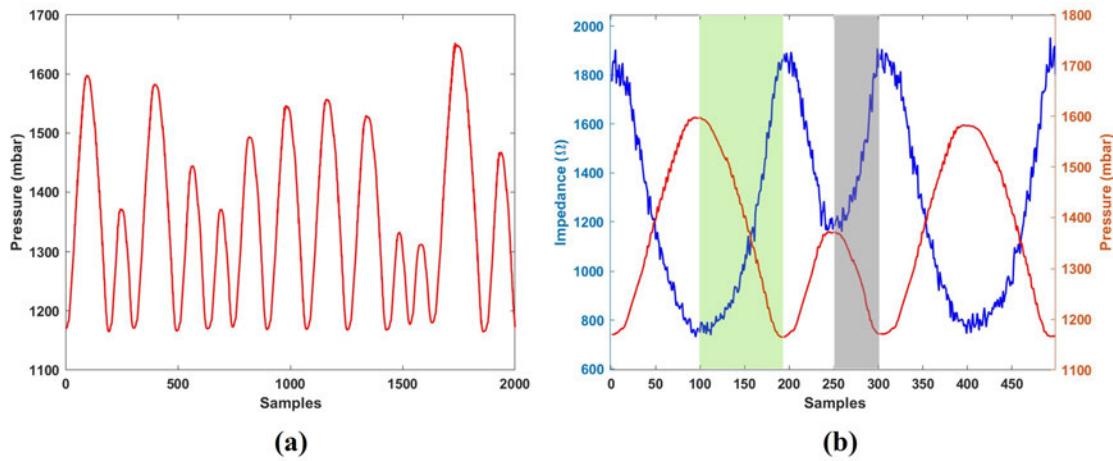


Figure 2.14: Pressure signal impact on EIT variations : (a) Random pressure signal applied for tip localisation data collection, (b) Pressure - EIT correlation indicates an indirect relationship in the shaded regions.

It is also instructive to examine how individual EIT channels respond under the eight-channel protocol defined in Fig. 2.7(b). To evaluate their behaviour, a pressure signal with constant amplitude was applied to minimise variability. Channels one, two, five, and six were selected for analysis (see Fig. 2.15). The first two channels were excited between electrodes two and 13, while the latter two were excited between electrodes three and twelve. The latter channels are positioned closer to the SCM's centre.

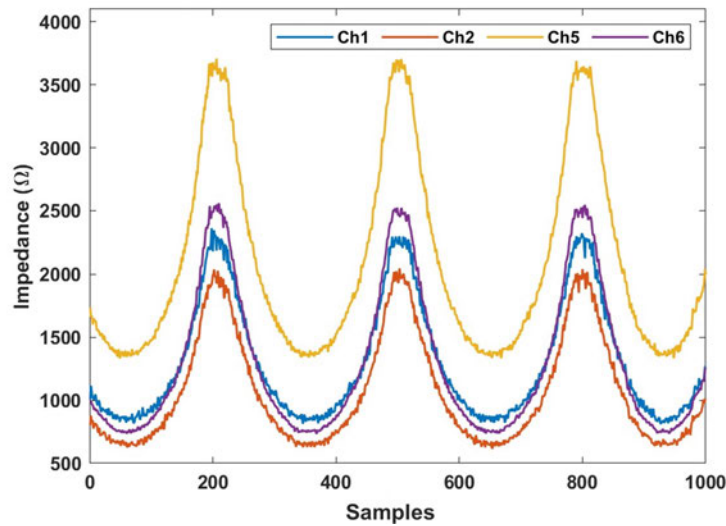


Figure 2.15: Comparison of EIT channel variations (channels one, two, five, and six) under a constant-amplitude pressure signal, according to the protocol shown in Fig. 2.7(b).

When comparing channels one and two, which share the same excitation electrodes, the difference between their peak-to-peak impedance (I_{pp}) is 1473.3 Ω for channel one and

1429.15 Ω for channel two. Although the difference is marginal, channel one exhibits slightly greater variation. The baseline impedance values (corresponding to the rest configuration with the smallest cross-section) are 2294.92 Ω and 2046.76 Ω for channels one and two, respectively. These results confirm that, as per the defined protocol, channel one measures a larger volume of saline, leading to higher impedance variations.

Similarly, for channels five and six, I_{pp} are 2360.78 Ω and 1806.61 Ω , respectively, with corresponding baseline values of 3699.65 Ω and 2558.17 Ω . Consistent with the earlier comparison, channel five exhibits more pronounced variations than channel six, owing to the larger conductive volume it covers. Comparing all channels, channels five and six show greater impedance variations than channels one and two, while also possessing higher baseline values. This indicates that the central region of the **SCM** experiences greater cross-sectional deformation and higher current density. In particular, since the current in channels five and six is injected through a smaller saline volume compared to channels one and two, the resulting impedance variations are amplified. The higher baseline impedance values in channels five and six further support this conclusion.

LSTM Performance

This section investigates the performance of **EIT**- and **LSTM**-based tip localisation and evaluates the effect of increasing the number of **EIT** channels. The prediction results for each channel configuration are presented in Fig. 2.16. Among the three configurations, using one, four, and eight channels, a qualitative assessment reveals that the predicted trajectories in both the y and z directions best match the ground truth when four **EIT** channels are employed for training and testing. The discrepancies between predicted and measured trajectories occur primarily at the peak points of the positional signals, with smaller deviations observed in the four-channel case. These results indicate that increasing the number of channels from one to four improves the prediction accuracy, whereas a further increase from four to eight channels leads to a reduction in performance.

A quantitative analysis using the **Root Mean Square Error (RMSE)** supports these observations. The minimum **RMSE** values for the tip position are 3.6 mm and 4.6 mm in the y and z coordinates, corresponding to 7.42% and 6.07% of the actuator's total range of motion (48.5 mm in y and 75.7 mm in z), respectively. These values are achieved by training the **LSTM** model with four **EIT** channels. The L_2 norm of the **RMSE** values in the y - z plane equals 6.24, 5.84, and 7.21 mm when using one, four, and eight channels, respectively. From the error plots in Fig. 2.16, the maximum errors in the y direction are 27.03, 17.5, and 26.31 mm for one, four, and eight channels, respectively, while the corresponding maximum errors in the z direction are 22.77, 18.62, and 22.7 mm.

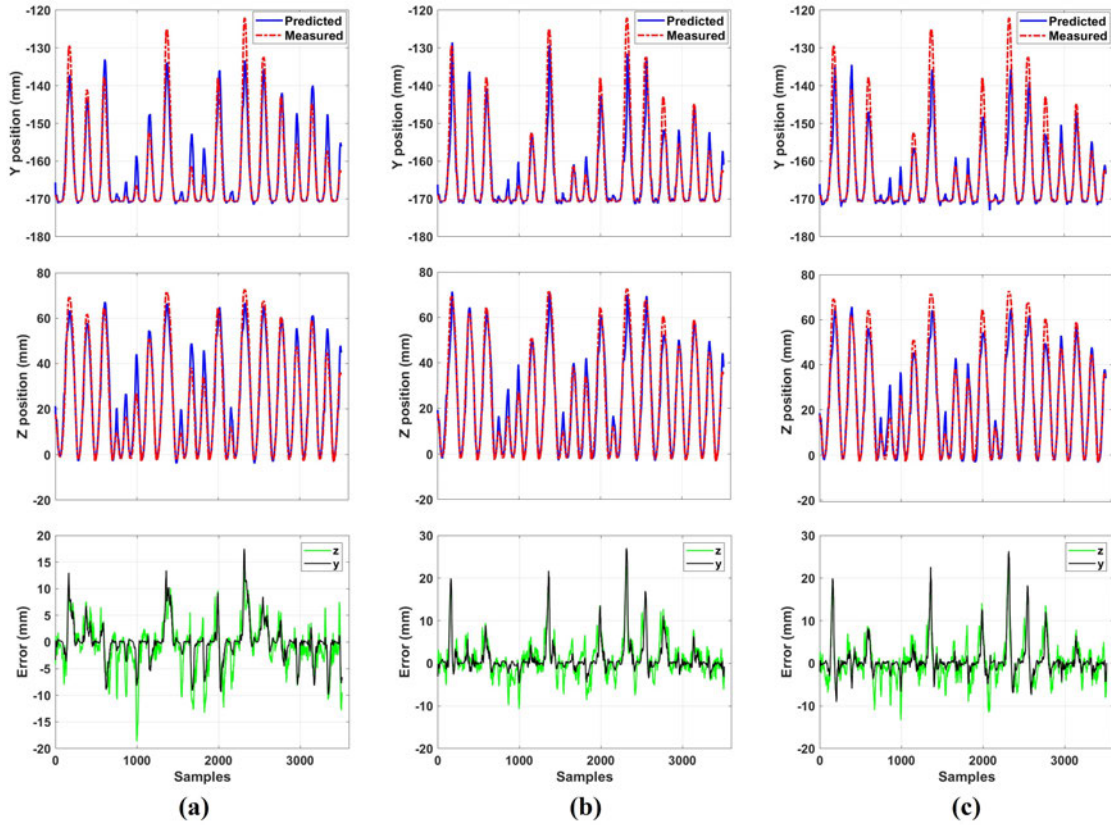


Figure 2.16: LSTM prediction results based on the number of EIT channels used for training: (a) One channel, (b) Four channels, (c) Eight Channels.

The range of motion and the predicted trajectories are shown in Fig. 2.17, where subfigure (b) illustrates that the four-channel configuration yields the closest agreement with the measured 2D trajectory. Both qualitative and quantitative analyses demonstrate that increasing the number of channels does not necessarily enhance prediction accuracy. Since EIT channels are not linearly independent, additional channels may contribute redundant information and introduce noise. Therefore, optimising the number of channels is crucial, as a larger number increases computational cost, wiring complexity, and manufacturing challenges without offering proportional improvements in localisation performance.

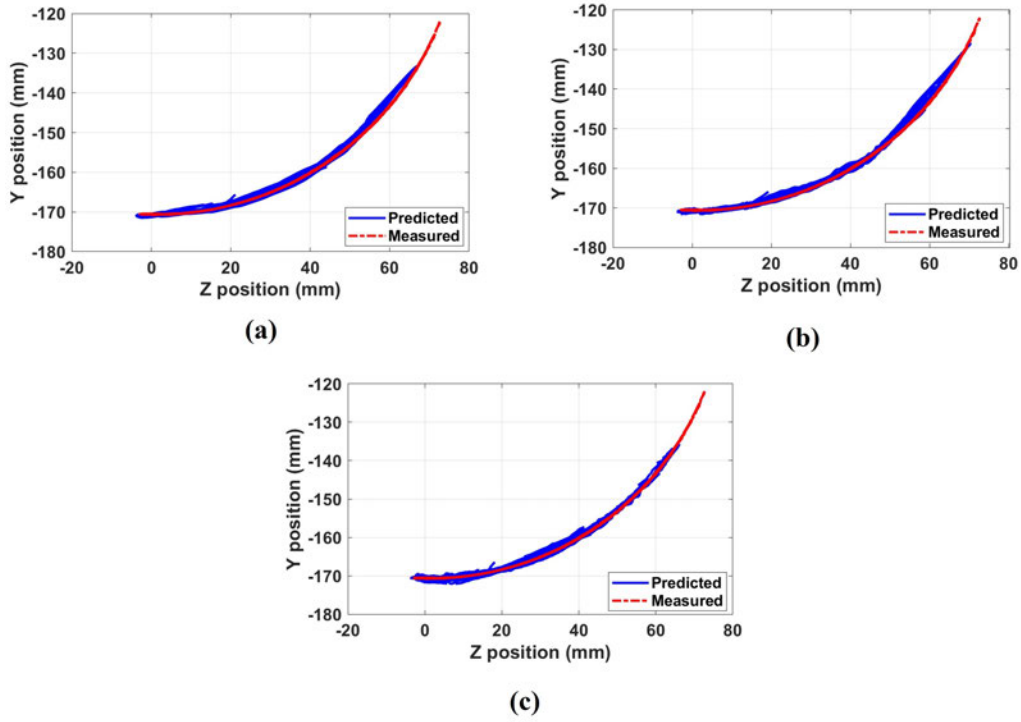


Figure 2.17: Ground truth and predicted tip trajectory through using different numbers of channels to train the LSTM model: (a) One channel, (b) Four channels, (c) Eight Channels.

2.4 Discussion

A promising strategy for designing a sensing scheme for SCMs hydraulically actuated by conductive fluids was investigated. This approach enables the development of a new generation of soft manipulators with accurate proprioceptive sensing, particularly suitable for endoscopic applications due to the biocompatibility, MRI compatibility, low cost, compactness, and scalability of EIT. In addition to these characteristics, both simulation and experimental evaluations demonstrated that EIT is reliable and robust, producing distinctive and proportionate outputs in response to varying deformations.

In this study, EIT was integrated into a continuum elastomer robot for the first time. The integration demonstrated a significant contribution to tip localisation in data-driven models based on LSTM networks, achieving a minimum RMSE of 3.6 mm and 4.6 mm in the y and z directions, respectively. When compared to similar studies on shape sensing in soft robots [152], which reported an average error of 2.17 ± 1.65 mm (equivalent to 5.42% of the range of motion), the RMSE achieved by EIT (6.07%) is comparable. Although [152] employed conductive Polydimethylsiloxane (cPDMS) sensors that exhibit better structural compatibility with soft bending robots, these sensors suffer from high nonlinearity, drift, and hysteresis, which compound the inherent time-varying effects of

the robot. Moreover, [EIT](#) allows for tunable spatial resolution by adding or removing electrodes or channels, which is particularly advantageous in tasks such as tissue interaction where higher resolution is desired.

As highlighted in the experimental methods section, the [EIT](#) output in simulation arises primarily from variations in the inter-electrode spacing imposed by the adopted geometric modelling. In contrast, experimental observations indicate a different underlying mechanism. Specifically, impedance variations are attributed to the radial expansion of the [SCM](#), as the fibre reinforcements do not fully constrain radial deformation and the silicone walls exhibit elastic compliance. Moreover, in the experimental setup, the integrated [FPC](#) is non-stretchable, thereby invalidating the space-dominated explanation observed in simulation. This discrepancy is evident in the results section through the comparison between [Fig. 2.10](#) and [Fig. 2.12](#). While simulations show an increase in voltage with bending, experimental results demonstrate a decrease in impedance, which is directly proportional to the measured voltage. This opposing trend arises because the inflation-induced expansion of the [SCM](#) cannot be captured within the *SolidWorks* environment, resulting in the neglect of cross-sectional deformation. In addition, the assumption of constant curvature within each segment further constrains the simulated response. Consequently, the [EIT](#) output obtained in simulation fails to reflect the experimental behaviour, indicating that the simplifying assumptions adopted in the simulation studies fail to capture the true physical response of the system. To address this limitation, future work should employ [FEM](#)-based platforms [[155](#)], such as [Simulation Open Framework Architecture \(SOFA\)](#) [[156](#)], and other numerical frameworks capable of modelling complex soft-body deformations. Furthermore, the current kinematic formulation should be replaced with more expressive models that account for curvature variation along each segment. Such approaches include rigid-body equivalent formulations [[143](#)], [PCC](#)-based methods, and other model-driven techniques [[157](#)], which offer improved fidelity for soft robot kinematic modelling.

The abovementioned resources for modelling soft robots more precisely would also facilitate the exploration of [EIT](#) performance under material nonlinearity and structural variations [[158](#)], enabling validation of tip position estimation in simulation prior to experimental trials. As experimental data collection is time-consuming, such simulation validation would offer substantial efficiency gains.

The tip localisation experiments revealed that varying the number of channels influences overall prediction accuracy when model hyperparameters are held constant. This finding aligns with the simulation results, where increasing the number of electrodes affected the output-to-configuration ratio. While using fewer channels may yield better results for free-motion localisation, in constrained scenarios, such as contact with tissue, using

a larger number of channels can provide richer deformation information, which is essential for enhanced proprioception. Consequently, optimising the number of channels or electrodes based on the operational context should be further investigated.

Although reduced accuracy was observed with eight channels, this cannot be solely attributed to channel count. Increasing the number of channels enriches the feature space, providing more information for shape variations. However, if the dataset size remains constant or insufficient for higher-dimensional input, model performance may deteriorate. This issue could be mitigated by collecting a larger volume of training data.

Beyond the number of electrodes and EIT channels, the configuration of the measurement protocol also plays a key role in tip localisation accuracy. As demonstrated, EIT variations were more pronounced in the middle section of the SCM, where deformations are greatest. Future studies could examine whether redefining the measurement protocol to focus on this region improves LSTM prediction performance.

Further exploration of EIT characteristics could provide deeper insight into its sensing capabilities. For instance, evaluating EIT sensitivity with respect to increasing bending angles would account for the nonlinear behaviour of silicone, whose expansion rate is also nonlinear. Long-term stability analysis of EIT outputs would also be valuable for identifying potential sensor drift, hence extending its applicability to extended operations. Since EIT readings are influenced by both the mechanical behaviour of the soft body and the electrical characteristics of the medium, understanding their respective contributions to signal quality is crucial.

To enhance LSTM-based tip localisation performance, several improvements can be implemented. In Fig. 2.15, the EIT data from all channels exhibited noticeable noise, which can increase estimation errors. One contributing factor is the presence of air bubbles in the working chamber, which must be carefully removed to ensure cleaner data acquisition. Another is the contact impedance between the saline and electrode pads, which can be mitigated by optimising electrode geometry [159]. Furthermore, the LSTM hyperparameters used in this study were selected heuristically; hence employing a systematic optimisation strategy could enhance performance. Benchmarking against alternative models with lower computational cost could also provide useful insights [160].

Future work will explore EIT performance in SCMs with multiple working chambers. While the single-DOF SCM presented in this chapter bends only in the vertical plane, extending it to a multi-chamber design would enable 3D motion, resembling the workspace of practical endoscopic or surgical instruments.

An important avenue for future research is investigating whether EIT can be utilised for exteroception as well as for proprioception. Due to their inherent compliance and adaptive morphology, soft robots can conform to external stimuli [161]. External forces induce deformations that propagate to the internal chambers, reflected by EIT through impedance variations. Thus, EIT could potentially detect external contact forces, enabling multi-modal sensing schemes relying solely on EIT for internal and external perception.

Chapter 3

EIT for Tactile Sensing

3.1 Introduction

The aim of this chapter is to investigate whether EIT can be employed for force estimation. Building upon the promising results presented in the previous chapter, where EIT successfully localised the tip position of a soft bending robot, it is compelling to further explore its potential for force estimation and tactile sensing applications. The intrinsic scalability of EIT, along with the validation of its performance in both exteroception and proprioception, can be leveraged to advance the development of intelligent surgical instruments with unified sensing capabilities, enabling the perception of both internal and external stimuli.

Estimating forces exerted by soft robotic colonoscopes is as critical as shape sensing for patient comfort [14]. Force feedback enables safe tissue manipulation during therapeutic interventions by preventing excessive loading, thereby minimising trauma [162]. This benefits both surgeons and patients: surgeons gain haptic feedback that enhances operational consistency and intuitiveness [163], while patients experience reduced complications and improved safety [164]. Haptic feedback has further been shown to increase accuracy, shorten procedure time, and improve success rates [165].

Force information also supports advanced control strategies [166], enabling higher levels of autonomy. In procedures such as ESD, knowledge of applied force is critical, as delicate tissue manipulation risks perforation if forces are excessive [167]. Monitoring applied forces provides a foundation for closed-loop control. Numerous studies have therefore investigated force prediction from soft robotic shape-sensing modules [168, 169, 131].

Beyond operational safety and comfort, force sensing has diagnostic potential. Many soft endoscopes rely on endoscopic cameras or intraoperative imaging for navigation and lesion detection [170], yet visual diagnosis struggles with early-stage cancer tissues as they remain invisible. Tactile sensing addresses this by detecting mechanical disparities, as malignant tissues often exhibit stiffness distinct from healthy tissue [10, 11, 171, 172]. Accordingly, soft tactile sensors for tissue palpation have been developed and categorised by their technological principles (Table 3.1) [173], employing indentation, catheterisation, or non-contact approaches [174].

Utilising touch for tissue manipulation enhances identification of lesions [175], achievable by soft tactile sensors with human-level accuracy [176] employed to detect the depth of hard inclusions within soft tissue [177, 178, 179, 180]. Conventional robot-assisted palpation, often with rigid manipulators, relies on force measurements during *ex vivo* operations, requiring tissue removal [181]. Another category of tactile sensors integrates with laparoscopic or endoscopic instruments, providing intra-operative tactile diagnosis without tissue removal [182]. Some use optical fibres to sense tissue stiffness [183, 184, 185], offering high resolution but increasing cost and complexity [173]. Additional tactile sensors measure the force exerted by miniaturised and articulated probes, providing information about tissue properties [186]. Chuang *et al.* developed a piezoelectric sensor for colonoscopies, mounted at the colonoscope tip, capable of detecting tissue stiffness variations [187]. Fluid-driven tactile sensors, enabling cost-effective and easy fabrication, identify tissue stiffness through volume changes [188] or fluidic channel impedance variations [189]. The sensors outlined in the literature either require tissue excision or depend on separate devices for mounting and operation.

In this chapter, the same LSTM architecture used in the previous chapter is implemented to estimate tip forces from the planar SCM. After validating the method's performance and repeatability, we extend its use to tissue characterisation by exploiting force estimation to distinguish stiffness variations. Therefore, we propose an EIT-based tactile sensing scheme integrated into a soft manipulator for colon navigation. This approach equips hydraulically actuated soft endoscopes with proprioceptive tactile feedback for tissue palpation, without additional sensing modalities, thereby enhancing their diagnostic potential for early-stage cancer detection. The method preserves the dexterity and compliance of soft endoscopes while minimising complexity and integration challenges.

Table 3.1: A summary of conventional tactile sensing technologies in MIS, including clinical feasibility considerations.

Sensing Technology	Advantages	Disadvantages	Clinical Feasibility	References
Piezoresistive/ Piezoelectric	- Low cost	- Prone to hysteresis	- Requires biocompatibility	[190]
	- High sensitivity	- Accuracy loss	- Medical safety concerns	[191]
	- Repeatable	in long-term		[186]
Capacitive	- Ease of design	- Prone to hysteresis	- Shielding requirement	[192]
	- Thermal noise immunity	- Non-linear	- Sterilisable material selection	[193]
	- Tunable resolution			
Optical	- Versatility			[194]
	- Miniaturisation	- Complex design	- Established use	[195]
	- Lightweight	- High cost	- Immune to EM interference	[184]
	- MRI compatible			[185]
Microfluidic/ fluidic	- High flexibility	- Leakage	- Sealing reliability	[196]
	- Low cost	- Limited stability	- Sterilisation compatibility	[189]
	- Easy fabrication	in long-term	- Miniaturisation	[197]
Imaging-Based	- High resolution	- Large-sized	- Ease of integration	[198]
	- Large area coverage	- Restricted camera	- Line of sight requirement	[199]
	- Minimised coverage	focal distance	- Additional processing hardware	[200]

The sensor operates by mapping impedance variations induced by tissue interaction, where the deformation of the continuum manipulator reflects the external force and tissue elasticity [12]. Since malignant tissues typically exhibit greater stiffness than healthy ones [11], their presence introduces distinct backbone configurations and bending curves during endoscope–tissue interaction (Fig. 3.1). Impedance measurements are captured through nine channels distributed along the in-plane bending manipulator in the previous chapter and processed with an LSTM model to classify early-stage malignant tissue samples on a phantom.

To validate this approach, a suspended phantom mimicking colon and mesentery tissue was developed, incorporating silicone samples of varying thickness to mimic stiffness changes. The soft manipulator was driven vertically against the phantom while recording pressure and impedance data sequences during interactions.

The author contributed to experimental design, hardware preparation, setup assembly, data collection, and analysis of the data-driven model. The Methods section details the experimental hardware, configurations, and acquisition procedure. The Results section presents dataset analysis and LSTM classifier performance, including colour-map visualisation of stiff tissue detection. Finally, the *Discussion* addresses limitations, suggests improvements, and outlines directions for more realistic experimental validation of this tactile sensing scheme. The contents and results of this chapter were partially adopted from [14], © 2023 IEEE, and [15].

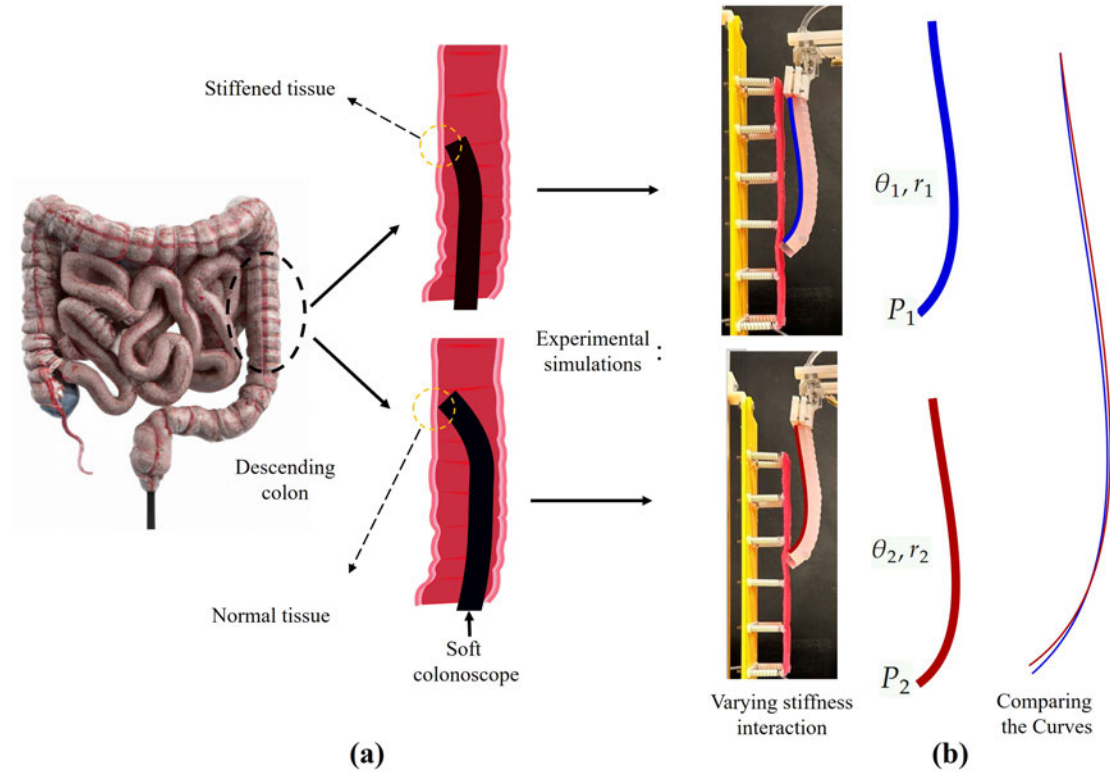


Figure 3.1: The schematics and experimental simulations of the proposed approach [15], which is based on the shape changes resulting from the interaction with different tissue stiffness: (a) Schematic illustration of a soft colonoscope sliding against the descending colon tissue; (b) Replicating the schematic representation in experimental simulations. Different shape configurations (e.g., bending angle and radius) over interaction with varying stiffness tissue samples.

3.2 Methods

3.2.1 Tip Force Estimation

The same **SCM** used in Chapter 2, shown in Fig. 2.4, was utilised to collect data for validating the force estimation capability of **EIT**. The **EIT** hardware, including the **FPC** (see Fig. 2.5(b)), impedance spectroscopy, control system (Fig. 2.6(a)), and the data acquisition protocols (Fig. 2.7(b)), were all configured as previously described in Chapter 2. The focus of the experiments in this section is solely on force estimation; therefore, the motion capture system used in earlier experiments was excluded from the setup.

In addition, the effect of the number of **EIT** channels on the accuracy of tip force estimation was investigated. The subsets of channels used for the one- and four-channel protocols are identical to those defined in the previous chapter. To measure the tip forces, a force/torque transducer (Nano 17, ATI Automation, USA) was positioned such that the **SCM** contacted it during pressurisation (see Fig. 3.2). The force sensor sampled data at a frequency of 62.5 Hz.

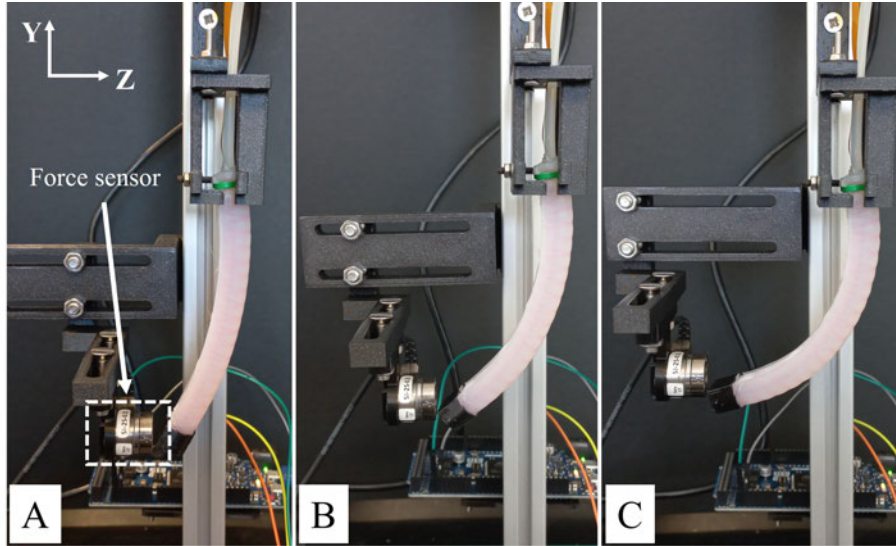


Figure 3.2: Collecting the contact force data for training the network in three different bending angles by pressurising the **SCM** with random amplitudes [14], © 2023 IEEE.

Data Collection

In addition to the effect of the number of channels, another variable factor was introduced as the location of the force transducer. The transducer was positioned at three different locations within the **SCM**'s working range to examine whether the configuration of the **SCM** at which the external tip force is applied affects the prediction accuracy. This design

was motivated by previous findings indicating that the tip localisation error increased with larger motion ranges.

Accordingly, the contact force dataset was collected at three different bending configurations of the SCM, denoted as **A**, **B**, and **C**, as shown in Fig. 3.2. In configuration **A**, the force sensor is closest to the SCM tip, while in configuration **C**, it is the furthest. The position of the sensor in the vertical (y - z) plane was adjusted using a rigid 3D-printed structure attached to an aluminium stand, allowing precise alignment along the tip trajectory.

Only the normal (z) component of the force vector applied to the transducer was considered for estimation, as it contains more significant force amplitudes. Forces acting out of the bending plane were assumed negligible, given the planar motion of the SCM. The SCM was actuated using a triangular pressure waveform with random amplitudes to enrich the dataset and capture a wide range of tip force values.

To ensure consistent contact between the SCM and the force sensor during pressurisation, the SCM was pre-bent so that its tip was at a minimal distance from the sensor before contact. This pre-bent configuration varied depending on the sensor position. Consequently, the minimum pressure levels in these experiments were higher than those in the straight, unpressurised configuration of the SCM. Specifically, the minimum pressures for configurations **A**, **B**, and **C** were 1104.5, 1400.2, and 1410.9 mbar, respectively. The corresponding maximum pressures were 1573.6, 1747.7, and 1753 mbar. It is evident that both the minimum and maximum pressure values increased as the force sensor was positioned further from the SCM tip (i.e., from **A** to **C**).

The data collection procedure yielded 49,664, 37,525, and 37,905 contact force samples for configurations **A**, **B**, and **C**, respectively. The applied force on the transducer acted along the negative z -axis. The absolute force magnitudes ranged from 0 to 0.111 N in **A**, 0.018 to 0.161 N in **B**, and 0.023 to 0.207 N in **C**.

Post-Processing

After data collection, the signals from the EIT, pressure sensor, and force transducer were synchronised and interpolated. The EIT and pressure signals were combined to form the input dataset for training the LSTM architecture shown in Fig. 2.8. The force data from the Nano 17 sensor in the z direction served as the ground truth and network output. The dataset was divided into training and testing sets with an 80:20 ratio. The same hyperparameters used for the LSTM network in the previous chapter were adopted here for tip force estimation.

3.2.2 Tissue Palpation

Colon Tissue Phantom Simulator

To replicate colon tissue stiffness for enabling EIT tactile sensing in characterising tissue stiffness, a set of phantoms is designed, considering the colon tissue suspension. Typically, phantoms overlook the normal stiffness of colon tissue, essential to mimic the resistance of ligaments and surrounding organs. The proposed phantoms in this chapter enable tactile sensory data collection. These phantoms, categorised into training and test phantoms, follow a mass-spring model. As shown in Fig. 3.3(b), an artificial tissue segment is suspended to a rigid component via compression springs with a constant of 175.1 N/m . Under quasi-static loading and a fixed spring constant, the elasticity modulus of 0.425 MPa is computed using the spring geometry (25 mm length, 6.5 and 1 mm outer and wire diameters, respectively), aligning with [201] results in the circumferential direction. The springs model the resistance yielded by ligaments and neighbouring tissues upon the palpation of the colon wall.

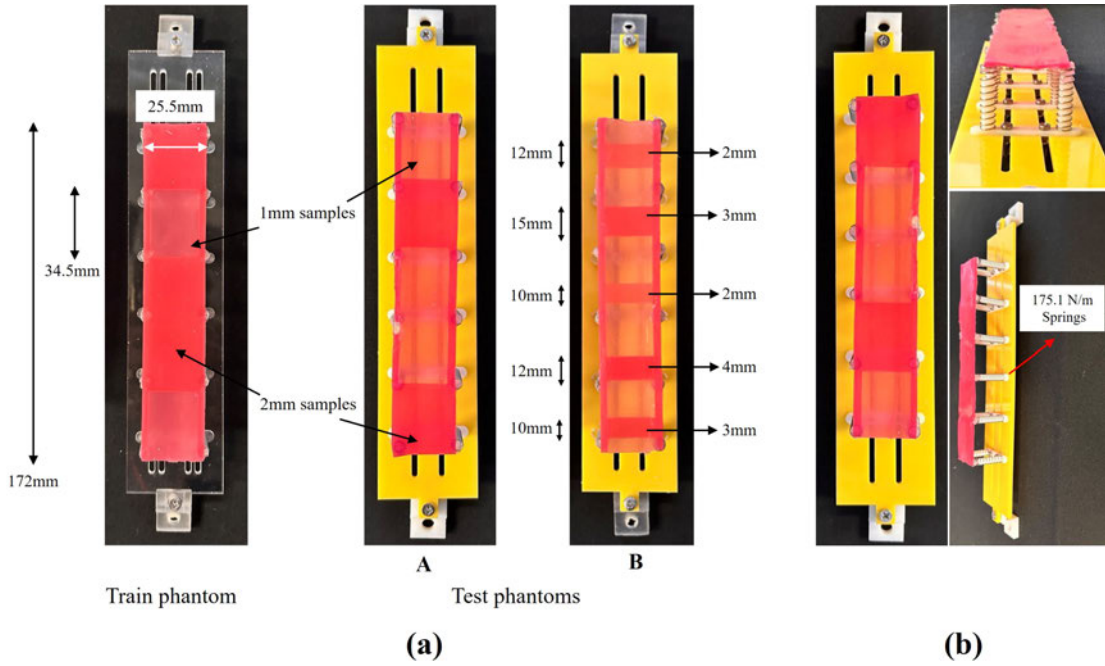


Figure 3.3: Features of the fabricated tissue phantoms: (a) The training and validation phantoms together with their specifications [15], (b) The structure of the mass-spring model and silicone samples of varying stiffness.

The tissue segment consists of Ecoflex 00-50 (Smooth-On, Inc., Macungie, PA, USA) silicone cast into 3D printed moulds, providing a location-based stiffness varying pattern through thickness variations. Stiffness variations are subsequently categorised through a binary classifier. The training phantom includes five tissue samples of 1mm and 2mm

thickness, with the latter indicating unhealthy tissues. The samples, measuring 34.5 mm in length and 25.5 mm in width, are randomly distributed across the segment. In Fig. 3.3(b), thicker tissue samples are identified by a darker colour. Test phantoms, A and B, follow similar fabrication principles, but A has an alternative distribution pattern of 1 mm and 2 mm samples, and B is composed of two additional samples of 3 and 4 mm thicknesses. While the lengths of the samples on phantom A are identical, they differ on B as depicted in Fig. 3.3(a). The varying design factors present in phantom B, namely sample length and thickness, serve as a means to investigate the generalisation of the proposed classifier. These tissue phantoms, combined with the soft manipulator, are employed to simulate colon palpation. A data-driven classification model will interpret tactile data collected during the procedure.

Experimental Setup

Tissue palpation involves probing the tissue while traversing the surface for data sampling at various locations. Two classes of actuation units are employed: one for probing motion and the other for the SCM's displacement in the vertical direction. Similar control unit in Chapter 2, Hydraulically pressurises the SCM and generates in-plane probing motion.

To probe the tissue, the manipulator tip indents the phantom, ensuring the exerted normal force stays within a safe range. The phantom is vertically mounted on an aluminium structure, and the manipulator is attached to the moving stage of a vertical linear motor, moving parallel to the phantom (see Fig. 3.4(b)). Stepper motors, controlled by S-lite and S series uStepper drivers, with a resolution of 400 steps per revolution, actuate the traversing and probing motions, respectively. Vertical manipulator displacements are recorded using a built-in encoder.

Prior to initialising the experiments, the SCM faces the tissue phantom with its flat side parallel to the phantom surface. To replicate *in vivo* trials, the manipulator's rest position is set within colon dimensions, ensuring its relative position to the phantom aligns with values from [202]. In Fig. 3.5 at $t = 0$, the manipulator is configured at its rest position with an absolute pressure of 1.260 bar. Pressure data were recorded throughout the experiments, transmitted to an Arduino Due at 20 Hz through an I2C interface.

The protocol employed for tissue palpation in Fig. 3.6 illustrates that three distinct pairs of excitation electrodes were used, each associated with three measurement pairs, totalling 9 channels. The modification from the previous protocol to the current one was made to ensure that the impedance variations are captured more uniformly across the SCM. Unlike the previous experiments, where deformations were primarily concentrated in the middle

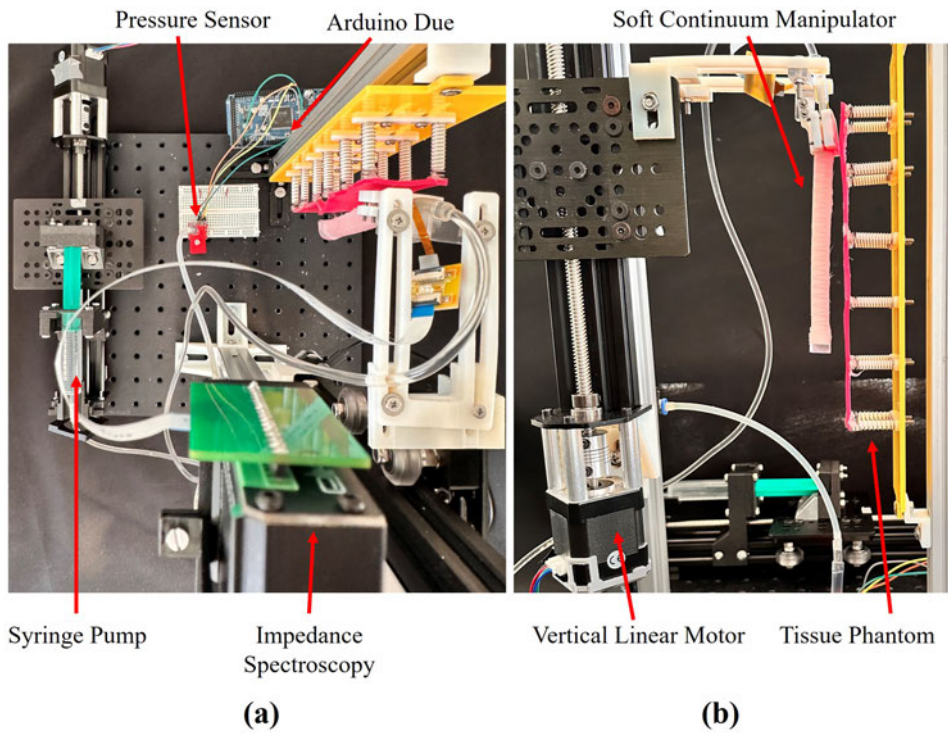


Figure 3.4: The experimental setup consisting of the tissue simulator, actuation units, continuum manipulator, and sensing devices [15]: (a) Top view, (b) Front view.

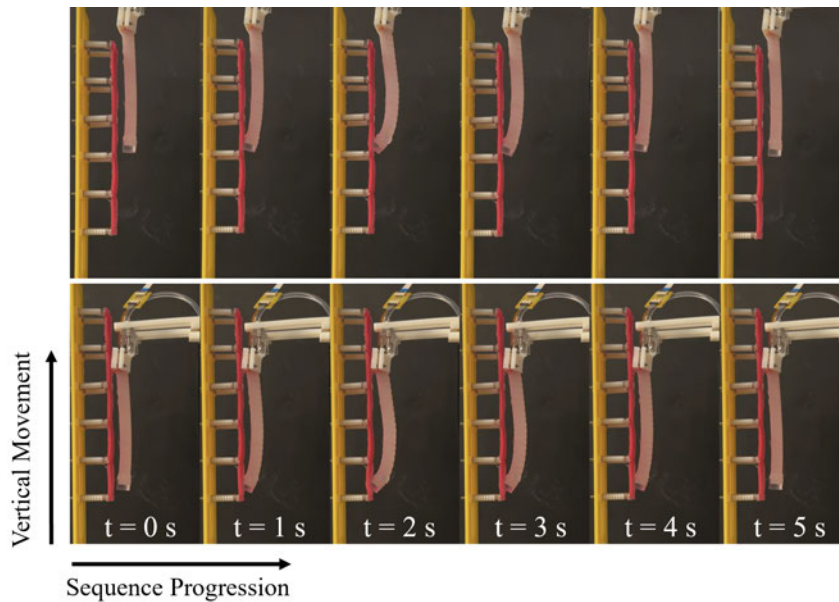


Figure 3.5: The inflation and deflation of the manipulator during data sampling, and the direction of traversing [15].

section of the [SCM](#), the presence of an external force during palpation may introduce distributed deflections along the entire body of the manipulator.

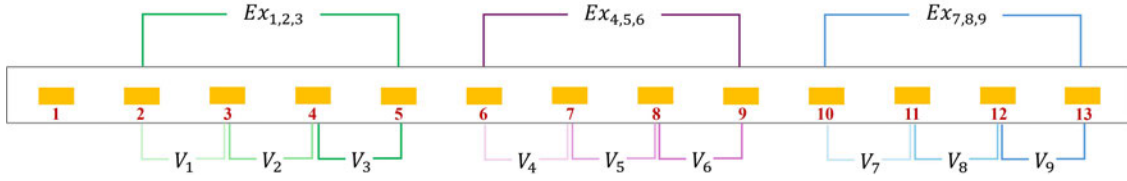


Figure 3.6: The EIT protocol with 9 channels, designated to capture impedance variations in tissue palpation experiments.

Data Collection

Two series of experiments were conducted. The first experiment demonstrates EIT output distinctions while probing tissue samples of varying thicknesses on the training phantom (Fig. 3.3(a)). The second experiment involves joint data collection on the training and test phantoms, followed by implementing a data-driven classifier. The experimental setup and required components for actuation and data acquisition are shown in Fig. 3.4. In the first experiment, pressure is applied to the manipulator in trapezoid signal waves with a constant high value of 1750 *mbar* and a high time of 2 seconds. The pressure is produced by the stepper motor rotating at the maximum speed of 2000 *steps/s* and acceleration of 6000 *steps/s*². This process is repeated at 5 different vertical positions of the manipulator tip, corresponding to distinctive parts of tissue samples with 1 and 2 mm thicknesses. These locations include the top, middle, and bottom sections of each sample. The experiment was repeated 10 times to assess the repeatability and separability of the EIT signals. In practical scenarios, factors such as the distance between the tip and tissue surface or the initial manipulator pose can vary randomly. Uncertainties involved during palpation due to the presence of complex colon anatomy and undesired motion of the manipulator render the surgical site unpredictable. This heightened uncertainty challenges the differentiation of tissue stiffness based on EIT outputs. To observe the impact of these uncertainties and create a more generalised dataset in the second experiment, the manipulator is pressurised with randomly bounded inputs from 1250 *mbar* to peak values ranging from 1600 to 1800 *mbar*. Despite the added complexity by randomising the pressure inputs, the overall robustness of the network is expected to be enhanced. Triangular wave signals of pressure (depicted in Fig. 3.13) inflate the manipulator instead of trapezoid ones. The bounds (1600 and 1800 *mbar*) ensure a safe interaction between the manipulator and the phantom while keeping the exerted force within a permissible threshold. The dataset from the second experiment comprises both training and testing data. The training data are acquired as the manipulator traverses and probes the training phantom (Fig. 3.5). Similarly, the test data come from palpating test phantoms A and B in Fig. 3.3(a). Both training and testing data collection start with the manipulator positioned at the bottom of the phantom and gradually moved to the top. The manipulator traverses the phantom at a

rate of 2 mm every 3 minutes during training, while the test data collection rate is set at 1 mm every 12 seconds.

Data Post-Processing

The data required for classifying palpated tissues include pressure values, EIT outputs, and the vertical displacement of the manipulator. Collected pressure and displacement datasets from Arduino and uStepper boards, respectively, are communicated to MATLAB 2022. These datasets, stored as string arrays with corresponding timestamps, are processed further. Impedance values, sampled at 20 Hz, are logged as text files and formatted into string and numerical arrays in post-processing. Pressure and EIT temporal sequences will be used as classifier inputs, while displacement data are used for labelling and illustration only. Labelling is based on the manipulator's position and relative vertical alignment to the phantom, considering tissue sample lengths and the manipulator's initial tip position. Each input sequence represents a single cycle of the SCM's inflation and deflation with corresponding EIT changes. EIT and pressure data are synchronised and interpolated using Piecewise Cubic Hermite Interpolating Polynomial (PCHIP) to ensure consistent data points in each sequence. Finally, the data are organised into train and test datasets, creating an n -by- d matrix for each sequence, where n represents the feature space size and d denotes the number of data points. The feature space size is ten (pressure and 9-channel EIT values) for all sequences, with d depending on the sequence's length.

3.2.3 Machine Learning Model

Due to the intrinsic nonlinear and hyperelastic nature of the SCM and tissue phantom, linear separability of data featuring different stiffness levels might be challenging. Colon tissue, in particular, exhibits isotropic and anisotropic hyperelastic biomechanics with time-dependent behaviour [203]. Thus, recognising nonlinear and temporal patterns in the data requires data-driven models for classifying colon tissue stiffness into healthy and cancerous categories. Similar to the previous chapter, an LSTM classification network is chosen as it effectively addresses the vanishing memory limitations of conventional RNNs while excelling at modelling temporal dependencies in sequential data of varying lengths.

For the tissue palpation dataset, a 6-layer LSTM network serves as the core algorithm for classifying the dataset into healthy and unhealthy tissues. Compared to the architecture highlighted in Fig. 2.8, the LSTM classifier employed in this chapter includes an additional softmax layer inserted between the fully connected layer and the output layer. Training data consists of 1057 sequences, reducing to 427 sequences for the test datasets on phantoms A and B. The input size, representing the feature space dimension, consists

of nine vectors for EIT channels and one pressure value. With a 2-sized output denoting healthy and unhealthy tissues, the network is configured accordingly.

Prior to inputting data into the classifier, the dataset is normalised and sorted based on sequence length to minimise the padding extent added by the network, thereby optimising the mini-batch size. The feature normalisation is performed using the maximum and minimum values of the datasets, resulting in scaling the feature variables to a range from 0 to 1. The LSTM classifier is trained using the *Adam* solver, and the execution environment is set to automatic. *Adam* enables faster convergence of the parameters by dynamically adapting the learning rate. The solver is suitable for noisy input data and robust to hyperparameter tuning. Training and validation of the network are implemented in MATLAB R2022a, and the learning algorithm is iterated over 500 epochs. The hyperparameters of the network including the number of hidden units, initial learn rate, and mini-batch size are tuned through the Bayesian optimisation algorithm [204], within the corresponding ranges of [1 100], [0.0001 0.1], and [1 300]. After 100 evaluations of the objective function defined as the error of classifications, the number of hidden units, initial learn rate, and mini-batch size are determined as 9, 0.02, and 170, respectively.

3.3 Results

3.3.1 Tip Force Estimation

Fig. 3.7 illustrates the correspondence between the input pressure, EIT response, and the resultant force (negative due to the direction of exertion) when the force sensor is positioned at location B. Comparing the pressure profile with that of the free motion case in Fig. 2.14 reveals notable differences, which are also reflected in the EIT signals across all four channels shown in Fig. 3.7(b). Specifically, the pressure signal here is considerably noisier, arising from intermittent contact and slipping events between the SCM tip and the force sensor. When pressure increases, the tip remains stationary until a threshold is reached, after which it slips, producing fluctuations in the pressure readings. However, these spikes are not directly mirrored in the EIT outputs. Another distinction is the slower rate of pressure rise and decay near local maxima, resulting from the tip's restricted motion due to the contact constraint. The correlation between EIT and pressure remains consistent with that observed in free motion, as seen across the four channels in Fig. 2.15. The corresponding force signal in Fig. 3.7(c) shows the exerted forces, with a maximum force of -0.16 N occurring at a proportionate pressure of 1747 mbar.

The estimation results across three EIT protocols and three sensor locations (Figs. 3.8–

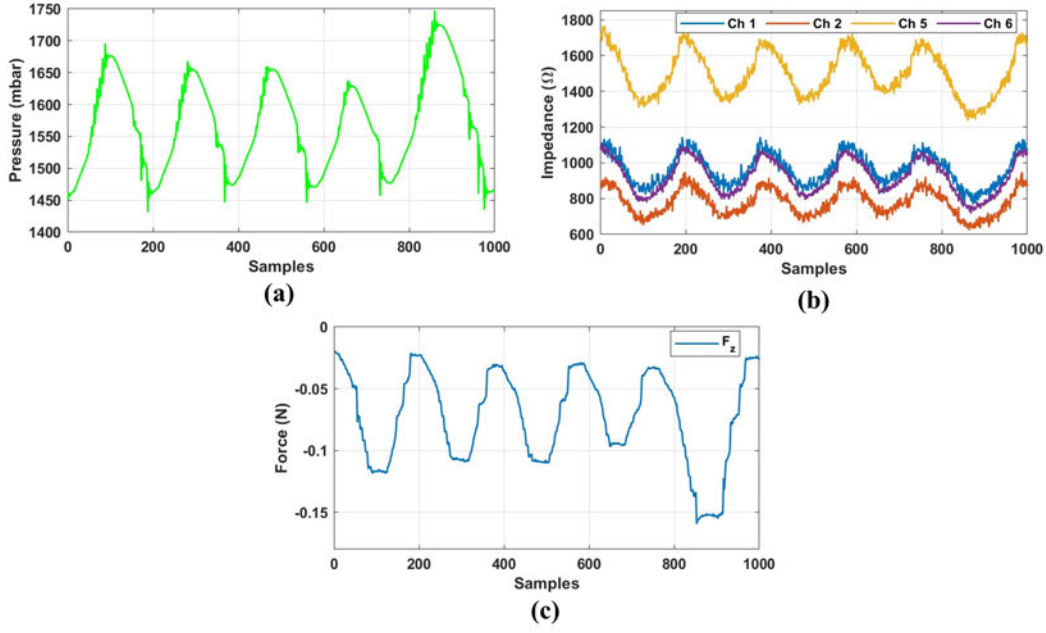


Figure 3.7: Raw data obtained during tip force estimation experiments from three sensing modalities: (a) Pressure, (b) EIT, (c) Force.

3.10) demonstrate that increasing the number of EIT channels consistently improves the force estimation performance of the LSTM model, as reflected in the corresponding error plots. Specifically, the single-channel configuration yields the highest error across all locations. Table 3.2 summarises the RMSE and standard deviation values for each configuration, while Fig. 3.11 presents these results in bar-plot form, where error bars denote SD and bar heights represent RMSE magnitude.

The lowest error occurs when the force sensor is located at position B (see Fig. 3.2 for reference) and the eight-channel EIT protocol is used, yielding an RMSE of 3.41 mN, corresponding to 2.11% of the maximum measured force. Conversely, the highest error is observed at position A with the one-channel configuration, resulting in an RMSE of 18.82 mN, equivalent to 16.95% of the maximum applied force. Overall, Fig. 3.11 shows that errors are largest at position A (closest to the SCM's straight configuration), followed by position C (farthest from the tip). The high inaccuracy at position A arises because the applied tip force generates a larger torque at the SCM base, causing bending localised near the fixed end rather than distributed along the body [205]. Since these proximal deformations are less reflected in the EIT readings, the estimation error increases. The higher errors at position C are consistent with those observed at trajectory maxima in Fig. 2.17, suggesting that positioning the force sensor near such regions similarly affects estimation accuracy.

Furthermore, regardless of sensor position, both the mean and SD of errors decrease with a

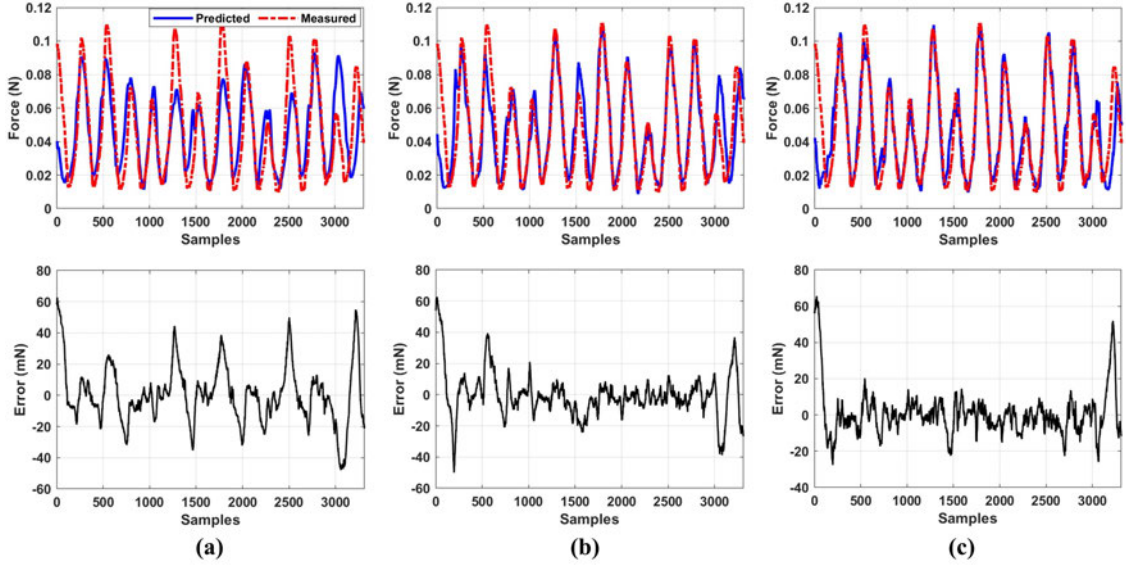


Figure 3.8: Force estimation performance of [EIT-LSTM](#), based on the [EIT](#) protocol with the force sensor located in A: (a) One channel, (b) Four channels, (c) Eight channels.

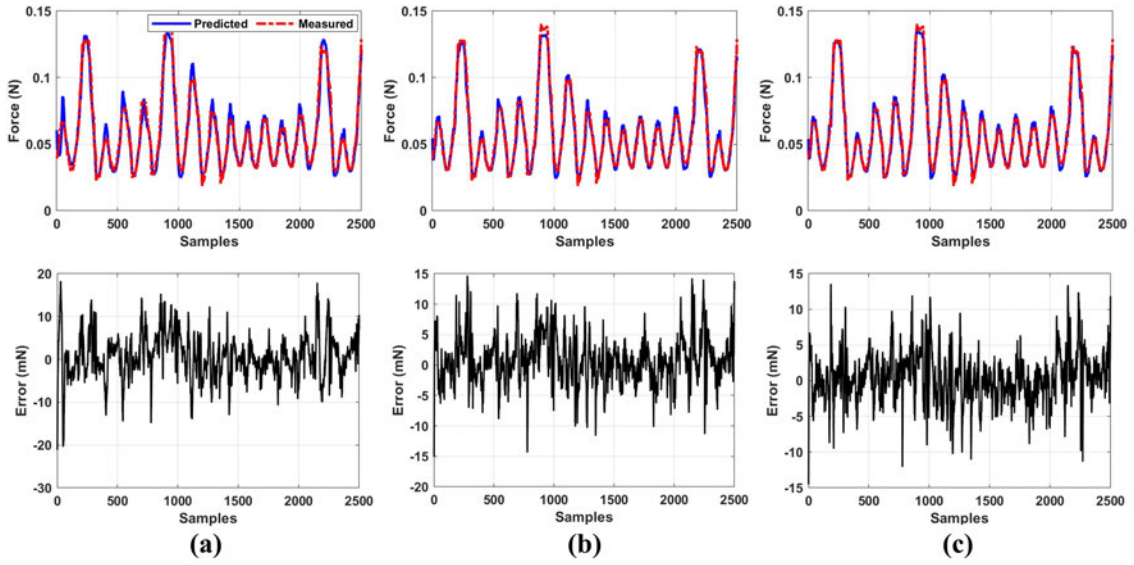


Figure 3.9: Force estimation performance of [EIT-LSTM](#), based on the [EIT](#) protocol with the force sensor located in B: (a) One channel, (b) Four channels, (c) Eight channels.

greater number of [EIT](#) channels, as additional channels provide richer spatial information on the actuator's internal state during force exertion. This finding contrasts with the results from the previous chapter, where increasing the number of channels did not improve tip localisation accuracy in free motion.

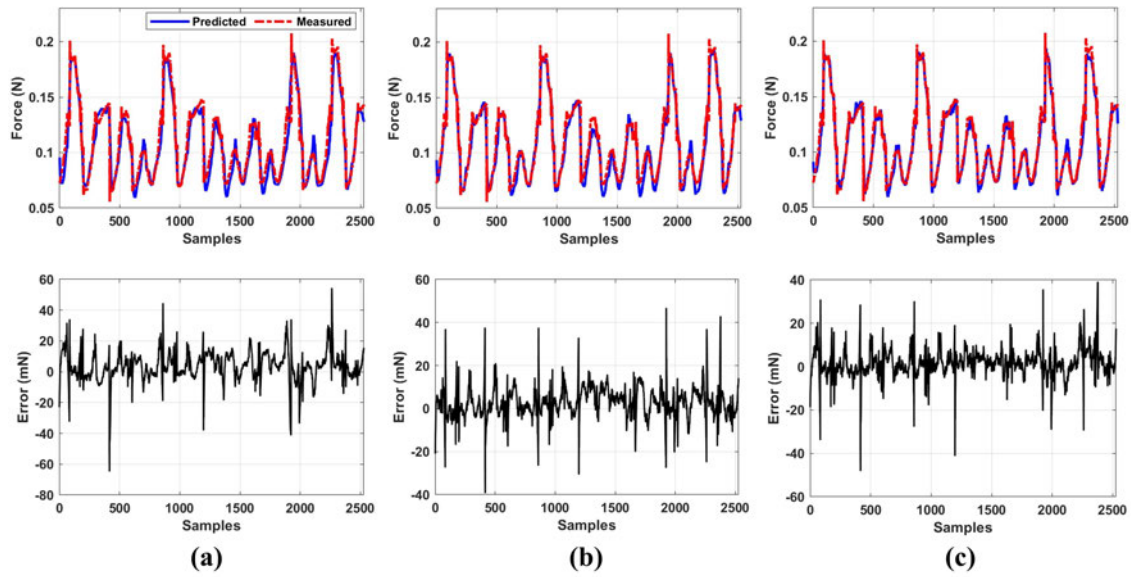


Figure 3.10: Force estimation performance of **EIT-LSTM**, based on the **EIT** protocol with the force sensor located in C: (a) One channel, (b) Four channels, (c) Eight channels.

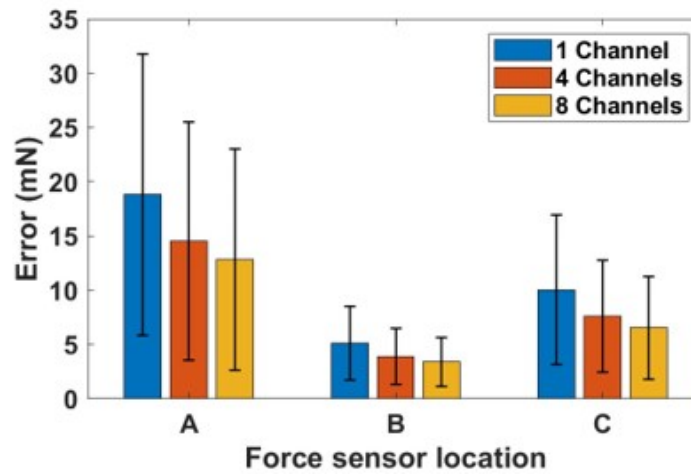


Figure 3.11: **RMSE** and **SD** of force estimation accuracy per location and number of channels.

Table 3.2: Force estimation errors (RMSE $\pm$ SD) in mN for different **EIT** protocol and sensor locations.

Location	One channel	Four channels	Eight channels
A	18.82 ± 12.98	14.51 ± 10.98	12.81 ± 10.18
B	5.11 ± 3.38	3.91 ± 2.60	3.41 ± 2.26
C	10.05 ± 6.91	7.59 ± 5.15	6.53 ± 4.74

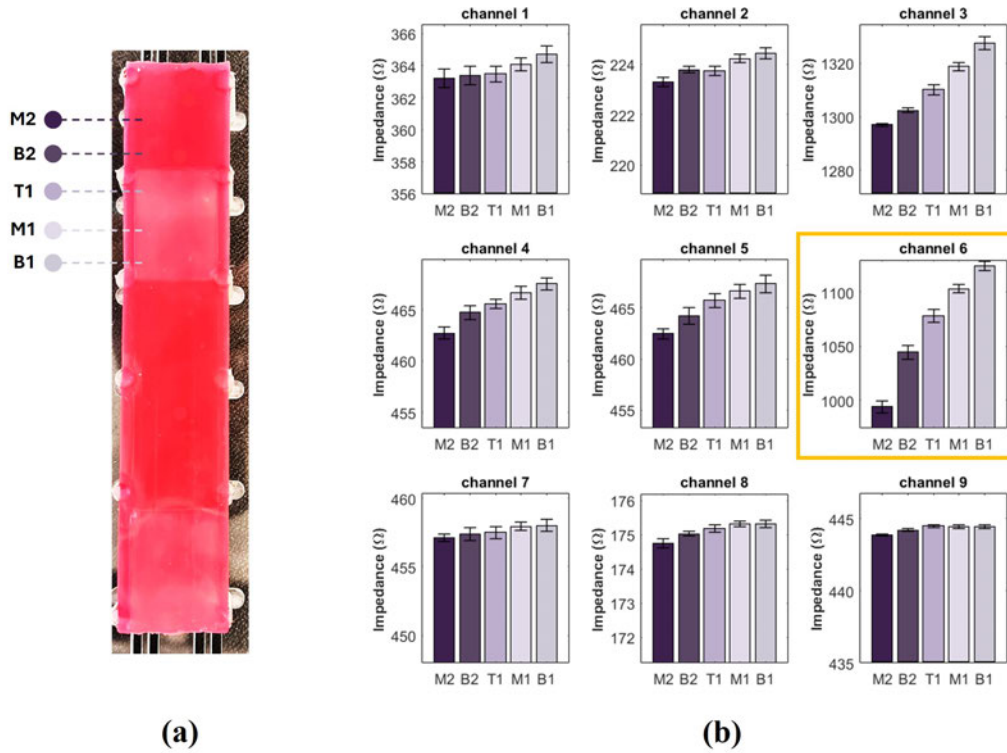


Figure 3.12: EIT results in palpation: (a) Regions with different stiffness on training phantom, colour-coded and labelled from M2 to B1 respective to their position on the phantom, (b) Bar plot of EIT outputs for all of the channels at their peak, showcasing the distinction of the outputs resulting from tissue interaction. Figure panels from [15].

3.3.2 Tissue Palpation: Experiment One

Fig. 3.12(a) labels tissue phantom samples as B1, B2, M1, M2, and T1, representing the bottom of 1 and 2 mm samples, the middle of 1 and 2 mm samples, and the top of 1 mm samples, respectively. The distinction is apparent in nearly all channels, with more pronounced variations in channels measuring impedances at the middle of the manipulator. This corresponds to the middle row framed in Fig. 3.12(b), aligning with electrodes in the middle of the manipulator per the experimental protocol. These high variations result from the middle section experiencing the most significant changes in cross-section. Additionally, the bar plots reveal that the distal section of the manipulator undergoes minimal changes in cross-section.

According to the plots, interaction with the middle of the 2 mm sample (M2 and B2) yields the lowest impedance value compared to other regions on the phantom. This suggests that cross-sectional changes are maximised when palpating the middle of the 2 mm sample, indicating the highest resistance in that interaction. This resistance is reflected in increased inflation across different sections of the manipulator. In contrast, interaction with the bottom of the 1 mm sample yields the highest impedance values, implying the

least resistance during palpation. The error bars depicted on the plots demonstrate the repeatability of the outputs. Overall, these values demonstrate the acceptable repeatability of EIT outputs.

3.3.3 Tissue Palpation: Experiment Two

A series of sequential data collected during the second experiment is shown in Fig. 3.13, where the pressure signal and corresponding changes in the EIT channels are demonstrated with respect to each channel. Individual sequences are separated by dashed lines. The sampled dataset inputs the LSTM classifier to train the model. The results of the validation using the data sampled from the test phantoms A and B are highlighted in this section. In particular, the data from the test phantoms are not present or involved in the training dataset and the learning algorithm, hence are considered unseen to the classifier.

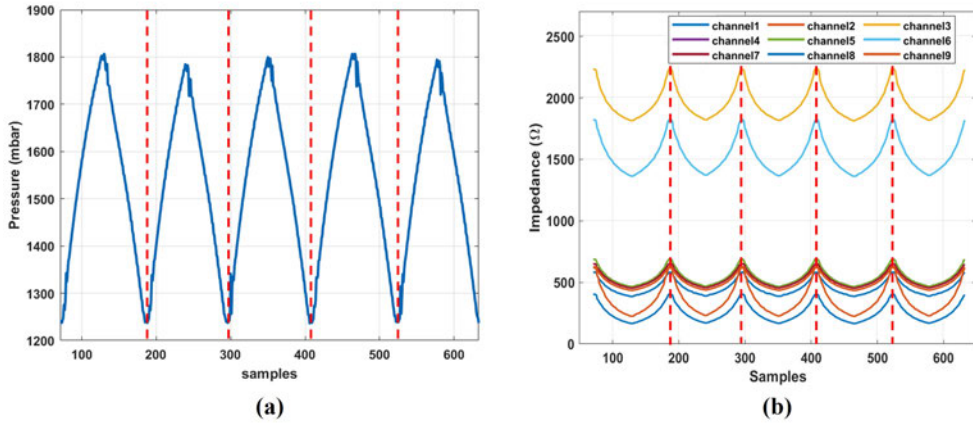


Figure 3.13: The sequential data obtained from the training phantom in experiment two: (a) Pressure signal, (b) Corresponding EIT changes in each channel, sequentially separated by dashed lines. Figures from [15].

The confusion matrix in Fig. 3.14(a) demonstrates the classification accuracy of the test dataset from the test phantom A. The phantom incorporates 1 and 2 mm samples, representing healthy and unhealthy tissue samples, respectively. The results indicate that the overall accuracy of the proposed classifier is 93%, implying the correct classification of 397 out of 427 sequences of the input data. The employed algorithms misclassified 30 sequences of healthy target samples as unhealthy tissues, resulting in an overall 7% false positive misclassification. In particular, 30 sequences of the sampled data from 1 mm samples on phantom A were classified as tissues with high stiffness.

Fig. 3.14(b) demonstrates the validation results of the proposed classifier using the dataset sampled from the tissue phantom B. Contrary to the test phantom A, phantom B incor-

porates tissue samples with 3 and 4 mm thicknesses in addition to 1 and 2 mm samples. Furthermore, the length of the samples is not constant, challenging the classifier more considerably. While the number of tissue samples on phantom A is 5, this number increases to 11 on phantom B. This value is associated with the number of transition regions where the thickness and stiffness of the tissue sample undergo a sudden change. Intuitively, the overall accuracy of the classifications drops to 88.1%, signifying the correct pattern recognition of 374 out of 427 sequential data samples. The remaining data samples are classified as false negatives, leading to the error of 11.9%. The confusion matrix illustrates that all of the data samples associated with healthy tissues, represented by 1 mm samples, are recognised correctly by the model. In contrast, the dataset representing the unhealthy tissue samples on phantom A was classified with 100% accuracy. The difference in the fabrication and structure of the phantoms as well as the sampling procedure can contribute to the varying accuracy between the datasets collected from phantoms A and B. In what follows, the respective locations of the misclassified sequences are highlighted using a heat map.

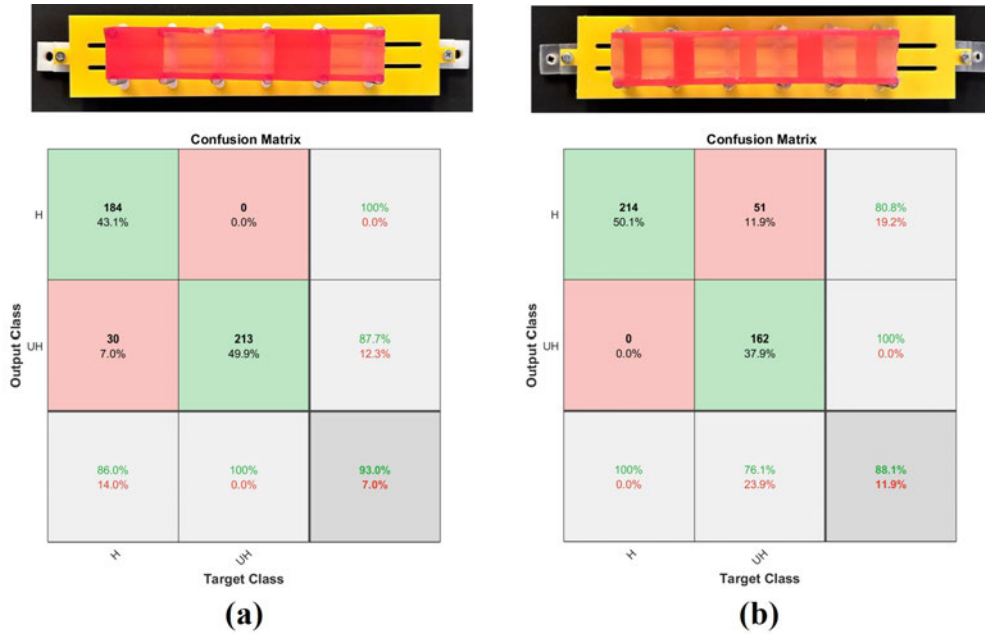


Figure 3.14: Classification results presented by confusion matrices with respect to the dataset collected on the test phantoms: (a) A and (b) B. Class **H** and **UH** denote the simulated healthy and unhealthy tissues with 1 mm and 2 mm thicknesses, respectively. Figures from [15].

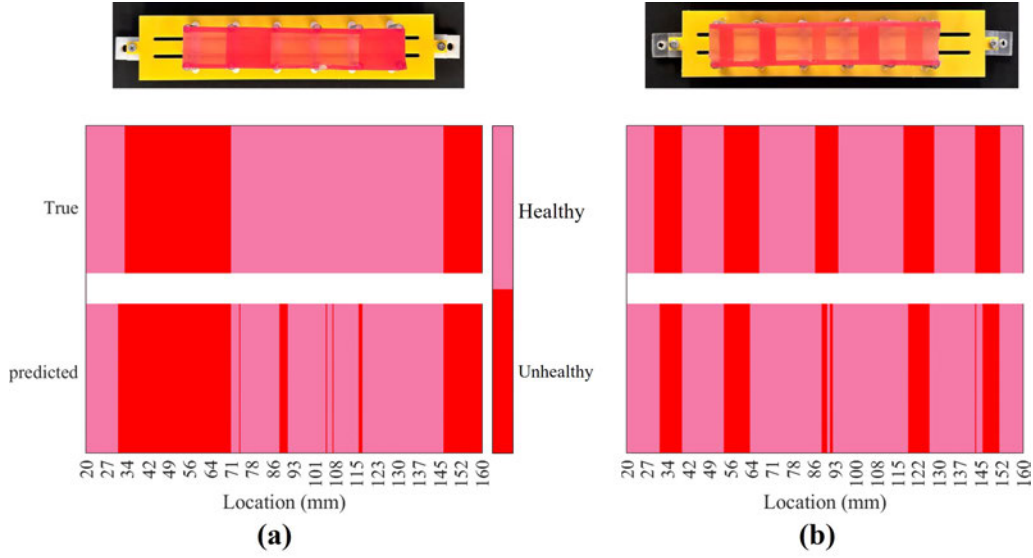


Figure 3.15: Heat map of the classification results for each phantom, indicating the location of misclassified data samples on the phantoms: (a) A and (b) B. Figures from [15].

3.3.4 Visualising the Location of Misclassified Data

Fig. 3.15 visually represents the locations of misclassified sequences on each phantom. Using the fabricated phantom as a reference, the figure illustrates two stiffness classes, visualised as colours with varying brightness. Bright colours represent healthy tissues or 1 mm samples, while darker colours indicate unhealthy tissues, including 2, 3, and 4 mm samples on both phantoms. The heatmap plots are divided into true and predicted visualisations on the vertical axis, denoting ground truth and classifier outputs, respectively. The true part aligns with the fabricated phantom, while the predicted part highlights misclassified palpations. The misclassification distribution on phantom A, as shown in Fig. 3.15(a), indicates that most misclassifications occur in the middle section. Additionally, some 1 mm samples between 27 to 34 mm are misclassified as unhealthy. Fig. 3.15(b) demonstrates that misclassified sequences on phantom B are concentrated within transition regions from 1 mm samples to higher stiffnesses. The stiffness changes within these regions are challenging to recognise through EIT outputs or the classifier due to identical stiffness properties. Errors in phantom B are expected at the boundaries between U and UH regions, avoiding false areas for investigation. However, in phantom A, a number of misclassified areas are rather scattered. To explain the error distribution pattern on phantom A, the anisotropic model of the phantom should be taken into account. The contributing factors include non-uniform moulding of the silicone tissue samples due to uneven distribution of the material during casting, and the relative position of the suspension springs that can vary the stiffness at particular areas. Additionally, since the manipulator traverses the phantoms in constant intervals over the collection of the test

data, an asynchronous vertical motion can induce misleading changes in the impedance values. In particular, if the manipulator is lifted while the tip is still contacting the phantom, the manipulator undergoes an unforeseen shift in its shape, impairing the classifier to recognise the input data. The limited response time of the proposed sensing scheme can amplify the effect of the aforementioned contributing factors.

3.4 Discussion

In this chapter, the force estimation capability of EIT combined with an LSTM network was explored and further extended towards medical applications such as tissue palpation. This approach aims to differentiate between soft and stiffened tissues, representing healthy and pathological conditions, respectively. Enabling palpation while traversing the colon provides an additional diagnostic advantage, utilising the same sensing modality required for shape estimation and thereby maintaining the overall structural simplicity of the endoscopic system.

The force estimation results indicated that the accuracy of prediction, given a similar LSTM model, depends strongly on the EIT recording protocol and the configuration at which the tip force is exerted. The most accurate results were achieved using an eight-channel EIT protocol with the force sensor positioned at location B (Fig. 3.2), yielding a relative error of 2.11% of the maximum applied force. This performance is comparable to, and slightly improved over, that reported in [152]. Conversely, location A with a one-channel EIT exhibited the poorest performance, achieving a relative error of 16.95%. This can be attributed to the intrinsic structural limitation of soft robots, which tend to bend primarily at the base with minimal deformation along other sections. As the internal pressure increases, the actuator's stiffness rises [206], making the unpressurised proximal end softer and more susceptible to torques generated by the applied tip force. In location A, where the torque arm is maximised, this phenomenon becomes particularly prominent.

Force estimation data collections can be repeated using stiffer silicone materials, such as Dragon Skin, to increase the measurable force range. While this change might also affect the EIT signal range, the enhanced material stiffness could improve measurement robustness.

In the present study, force estimation was limited to one direction (the Z-axis) and applied only at the tip. Future work should extend this framework to three-dimensional force estimation and include multiple contact points along the SCM. Ultimately, this would enable simultaneous estimation of both the direction and magnitude of the applied force.

Given the accuracy of the eight-channel protocol in estimating contact forces, the tactile sensing capability of EIT was further utilised for tissue palpation. The approach holds potential for facilitating tissue palpation during endoscopy, allowing for in situ localisation and assessment of tissue stiffness. Experimental investigations conducted on simulated tissue, combined with a ML classification method. The distinction between healthy and unhealthy tissue within the phantom is within a 1 mm difference in thickness, which challenges the classifier. The presence of contact resistance at the FPC electrodes contributes to the signal noise. Furthermore, the limited deflections created on the phantoms diminished the differentiability of measurements obtained from distinct regions. This limitation can be addressed by adjusting the stiffness of the manipulator and deploying variable stiffness designs in future work. The results demonstrated the achievement of 93% and 88.1% precision in classifying phantoms with 2 and 4 varying stiffness levels, respectively. Notably, previous studies by [180] and [181] reported accuracy rates of 97% and 99% in anomaly detection using robot-assisted tactile methods. However, these studies employed commercially established sensing modules, partially composed of rigid components and integrable with laparoscopic instruments. In contrast to the soft tactile sensors in the literature [207, 208, 73], the proposed proprioceptive tactile system in this chapter exclusively utilises the manipulator's pressurising fluid, without requiring supplementary attachments on the device's exterior. This feature preserves the structural stiffness and conserves space for the inclusion of interventional instruments alongside the endoscope.

Apart from the promising results yielded by the proposed algorithm and experimental setup, different aspects of the study require further improvement, which will be regarded in future works. Similar to the tactile studies in endoscopy [209, 196, 210, 198, 197], the fabricated phantom discussed in Section 3.2.2 represents the descending colon in a simplistic manner and lacks the corresponding physiological features such as the segmented appearance of the colon or haustra. While the mass-spring phantom possesses a flat surface, the majority of the existing phantoms in the literature consider haustra. However, these phantoms fail to simulate the mechanical properties of the colon such as stiffness, which is the key feature to identify cancerous tissues with palpation. Therefore, designing realistic phantoms that are identical to the large intestine enables accurate and generalised data acquisition with regard to *ex vivo* and *in vivo* colon tissue palpation [211]. The promising classification results on unseen tissue samples suggest that the proposed approach holds potential reliability when tested on real tissue in future studies.

Despite the qualitative correspondence between the elasticity of the designed tissue phantom and the study reported in [201], further calibration against the mechanical properties of real colonic tissue is required in future work. Established protocols exist for the

quantitative characterisation of soft tissue mechanics, including measurement of Young's modulus, shear modulus, and viscoelastic behaviour. These properties can be obtained using indentation tests, uniaxial compression or tension tests, rheometry, and elastography. Once the target tissue properties are quantified, calibration of the phantom can be achieved through an iterative fabrication process, in which the base material, structural configuration, and spring stiffness parameters of the mass–spring model are systematically adjusted until the measured response matches that of real tissue. Similar calibration strategies have been reported in the literature, for example using uniaxial mechanical testing [212] or elastography-based validation, and can inform future phantom development.

The in-plane bending of the manipulator limits its motion to one DOF, inhibiting the validation of the proposed method in bio-realistic 3D phantoms. To enable thorough palpation of the colon and considering the limited motion of the manipulator, a rotational actuation unit is required to complement the vertical motion, complicating the overall delivery system design. In addition to the need for a manipulator capable of 3D movement, enhancing the delivery system to convey the probing manipulator to the targeted tissue is required. Here, a vertical linear motor is leveraged to enable the manipulator to traverse the tissue surface. To develop a targeted delivery system and enable palpating tissues further inside the intestine, soft everting robots or alternative everting mechanisms [213] can be employed. These systems are capable of compliantly and flexibly conveying the palpating device to the targeted region. Additionally, the palpating manipulator can be mounted on a conventional endoscope delivering the treatment to the surgical site. In this chapter, controlled motions with random depths were employed; however, the direct implementation of these motions is limited in clinical settings. To generalise the ML model in clinical scenarios, it is necessary to incorporate data acquired from previous trials into the training dataset. Furthermore, the deployment of support structures, such as inflatable scaffolds with variable stiffness [214], can effectively enlarge the surgical site. Additional anchoring mechanisms, such as double-balloon systems [215], can contribute to the enlargement of the workspace and mitigate external forces. The integration of an additional sensing channel into the SCM can capture readings associated with the random motion of the endoscope, enabling the distinction of tactile measurements.

In the tissue palpation experiments, only normal forces along the z -axis were considered significant to create indentation and hence affecting the shape of the SCM. This simplification was primarily adopted for proof-of-concept validation. However, under clinical conditions, the instrument is subjected to complex force vectors comprising both normal and shear components. In addition, tangential interaction forces, for example those arising from friction, are continuously applied along the entire length of the SCM and may

challenge sensor robustness. Future work should therefore incorporate multi-axis loading conditions into the experimental design, enabling the validation of the proposed approach under clinically relevant scenarios and supporting combined force estimation using EIT.

The impedance spectroscopy records and exports EIT measurements through text files, which demand further data post-processing, impairing the real-time classification of the collected data samples. To address the issue, the current EIT system, which operates based on TDM and requires constant electrode switching, can be upgraded to a Frequency Division Multiplexing (FDM) based acquisition unit [8]. The current setup limits the real-time acquisition and processing of the data samples, induced by the limited sampling rate in TDM. Introducing FDM-based acquisition removes the need to post-process the data, enabling on-demand classification.

In addition to the employment of a TDM-based EIT system, the proposed classifier based on RNN can impose constraints on the real-time classification of the datasets. Although RNN and LSTM algorithms surpass CNN and DNNs in processing temporal data, RNN-based algorithms rely on a recursive mechanism, increasing the training time of the model. Transformer networks predict sequential data following a self-attention approach, where the attention and dependency between data points in a sequence are captured more effectively. This procedure is performed by processing the sequential data as a whole rather than point-wise, which is opposed to RNNs [216]. However, transformers commonly struggle with overfitting and generalisation, introducing data-augmentation techniques as a solution [217]. Transformers have demonstrated prominent results in modelling of soft robotics sensory data as in shape reconstruction applications [58], rendering them promising algorithms to interpret tactile data. Additionally, implementing a model with prediction uncertainty, such as a probabilistic model, could provide insights into the model's confidence levels regarding its classifications. This approach may offer valuable information about boundaries and potential misdiagnoses, particularly in the middle section of phantom A.

Implementation of data-driven classification methods poses challenges regarding data-collection time. In particular, one of the advantages of elastomer robots in endoscopic applications is their cost-effective manufacturing, offering the development of disposable surgical tools. The disposability of soft robots is achieved by repeatable manufacturing processes, allowing fabrication of robots with identical characteristics. However, the conventional fabrication approaches, typically using silicone casting, pose challenges in fabricating repeatable elastomer endoscopes [17], requiring larger training datasets to ensure generalisation, hence extending data collection time. The issue can be addressed either by devising repeatable manufacturing processes or incorporating augmented data

obtained from simulation experiments. Alternatively, the design of bio-realistic phantoms together with devising efficient and time-saving pipelines for data acquisition suggests a potential solution to the matter.

An additional factor influencing classification accuracy is assumed to be electrode contact impedance with saline. Improvements can be performed by electrode coating techniques using biocompatible and conductive substances [218]. Furthermore, optimising the injection-measurement protocol contributes to minimising the training time and improving the accuracy. The integration of a higher number of electrode pairs in the proximal end and middle section of the **SCM**, optimised through simulations, can reflect the changes in the shape more effectively. Additionally, the material used to fabricate the manipulator can impact the sensitivity of the approach. In particular, softer materials undergo more significant cross-sectional changes upon pressurising, hence exhibiting more sensitivity when interacting with tissues of varying stiffness.

The use of a single excitation frequency for **EIT**-based tissue classification limits its suitability for hybrid force–shape estimation in future applications. In contrast, **EIS** can play a key role by enabling analysis of impedance responses across a frequency spectrum, thereby providing access to complementary mechanical and electrochemical information. In **EIS**, impedance can be decomposed into charge-transfer resistance (R_{ct}), double-layer capacitance (C_{dl}), and Warburg impedance (Z_w). This decomposition enables the effects of applied force and internal pressure on the measured response to be decoupled, thereby supporting future investigation of hybrid estimation strategies. R_{ct} characterises the electrochemical reaction kinetics at the electrode-electrolyte interface and predominantly influences the mid-frequency response [219]. The double-layer capacitance, C_{dl} , arises from charge accumulation at the electrode–solution interface and is most influential at low frequencies, where it reflects interface-area variations associated with cross-sectional changes in the **SCM** due to pressure or contact. The Warburg impedance, Z_w , dominates the low-frequency regime and represents diffusion-controlled mass transport of ionic species, which can be exploited to identify elongation of the **SCM** and changes in ionic transport pathways.

Following the promising results achieved in equipping the planar **SCM** with shape sensing and force estimation using **EIT**, it is essential to extend the applicability of **EIT** to spatial **SCM** configurations, where higher manoeuvrability enables more realistic trajectories and motions. It is important to investigate whether increasing the **DOF** impacts the capability of the proposed approach to deliver reliable outcomes. Upgrading to a spatial **SCM** is intrinsically accompanied by additional working chambers, **FPCs**, and electrodes, as well as the coupling behaviour among the chambers. The next chapter highlights how these

features determine the accuracy of **EIT**-based state estimation.

Chapter 4

EIT for Soft Manipulators in 3D Space

4.1 Introduction

This chapter presents the integration of EIT into a hydraulically driven SCM with the capability to operate in 3D space, in contrast to the manipulator employed in previous chapters to demonstrate the capabilities of EIT. Having established the effectiveness and reliability of the sensing modality in planar bending, the transition to 3D manipulation enhances dexterity, manoeuvrability, and target reachability within 3D environments. From an application standpoint, motion in 3D enables colonoscopy instruments to perform more complex triangulation and navigate anatomical tracts seamlessly compared to the planar scenario.

In colonoscopy, it is common that due to patient positioning or the inherent spatial configuration of the colon, the gastroenterologist must steer the colonoscope in multiple directions without rotating the entire instrument to orient the bending tip towards the target. Furthermore, due to the adaptability of EIT, there is potential for its integration into potential state-of-the-art, hydraulically driven soft surgical robots functioning in 3D environments. Therefore, this chapter elaborates on the integration of EIT into a manipulator capable of operating in 3D space. Given its low cost and computational efficiency, EIT may be incorporated into SCMs with more than three working chambers and those with higher DOF. Nonetheless, the scope of this study is confined to validating the approach within a three-chambered SCM. The primary contribution of this chapter is to assess EIT performance in capturing the shape changes of the SCM in free motion. The design and optimisation of the FPC layout, hardware development, control system implementation for data acquisition were all performed and validated, before experiments in 3D environments.

The proposed manipulator incorporates multiple working chambers, allowing for saline pressurisation and consequently enabling control in three bending planes, rendering it a 2 DOF system. The fabrication procedure follows the same principles described in Chapter 2, namely, the design of appropriate 3D-printed moulds followed by silicone casting. The mould and manipulator design are inspired by the STIFF-FLOP device in [220] and [221, 222], albeit with modified dimensions to suit the intended application. The manipulator specifications were determined iteratively, in preliminary experiments, to ensure suitability for evaluation in pre-clinical and clinical colonoscopy trials. Design criteria included the outer diameter, total active length, dimensions of the working chambers, and wall thickness. The mould accommodates additional space for the potential integration of surgical instruments or vision systems. The silicone selected for casting the SCM possesses greater hardness than the EcoFlex 00-50 used in earlier chapters. This material choice was motivated by two factors: (1) the transition towards more realistic SCMs suitable for pre-clinical testing, necessitating higher load-bearing capacity to counteract interaction forces; and (2) the planned integration of the developed EIT system into a 6 DOF manipulator with three segments [222], which was initially fabricated using the same material.

Unlike previous chapters, the FPC substrate design was revised to accommodate the complex, compound deformations experienced in 3D manipulation. Additional evaluations were undertaken to assess the robustness and reliability of the FPC output under expected deformation patterns in the embedded working chambers. The protocol defined for the 2 DOF manipulator is presented, informed by the findings on electrode number effects from Chapter 2, and constrained by hardware limitations on EIT signal acquisition.

Following the integration of the embedded system, a series of experiments was conducted to evaluate overall system performance. These evaluations encompassed hysteresis, flexibility variations resulting from the addition of the FPC, and signal repeatability. The hardware system employed to actuate the manipulator and sense multimodal data, including pressure and positional information, is also described. This setup builds upon the hardware used in planar SCMs, incorporating duplicated components along with newly added boards and drivers.

Subsequent data collection experiments involved acquiring both positional and proprioceptive data for use in the estimator model introduced later in the chapter. Two distinct pressurisation patterns were applied: (1) single-chamber pressurisation, in which one chamber is activated while the remaining two remain passive, and (2) multi-chamber pressurisation, involving randomised activation of two chambers in terms of order, timing, and amplitude to capture compound effects.

Upon successful EIT integration and sufficient data acquisition, data-driven inference of system output was pursued. As in the preceding chapters, an NN model, e.g., RNN, was trained and tested on the collected dataset. The network architecture in this chapter offers computational efficiency and low latency, allowing real-time feedback generation for closed-loop control, unlike RNNs with higher computation cost [223]. In previous chapters, due to the simpler, single-chambered actuator, the input-output dimensionality was lower and computational burden was minimal, rendering the use of RNNs justifiable. MLPs and FFNs have been extensively utilised for shape reconstruction tasks [224, 207]. Additional to their computational efficiency and suitability for real-time implementation, MLPs are considered a suitable choice for generalisation as well [160]. The following sections detail the network architecture and hyperparameter configurations. Finally, the reconstruction accuracy is assessed through evaluation of training and validation loss, as well as the fidelity of the predicted shape with respect to the ground truth, as measured by RMSE.

4.2 Soft Continuum Colonoscope Fabrication

To enable the validation of EIT under preclinical and more clinically relevant scenarios, it is essential to deploy the complete system as a colonoscope prototype. This configuration facilitates insertion and enables proper management of the tubes and cables associated with the embedded sensing system, thereby enhancing the resemblance to a conventional colonoscope. This section details the development of the Soft Continuum Colonoscope (SCC). The fabrication process comprises three main stages: silicone moulding of the SCM, integration of the FPC, adding the soft caps to seal the SCM, and finally the addition of a passive, flexible overtube to facilitate insertion.

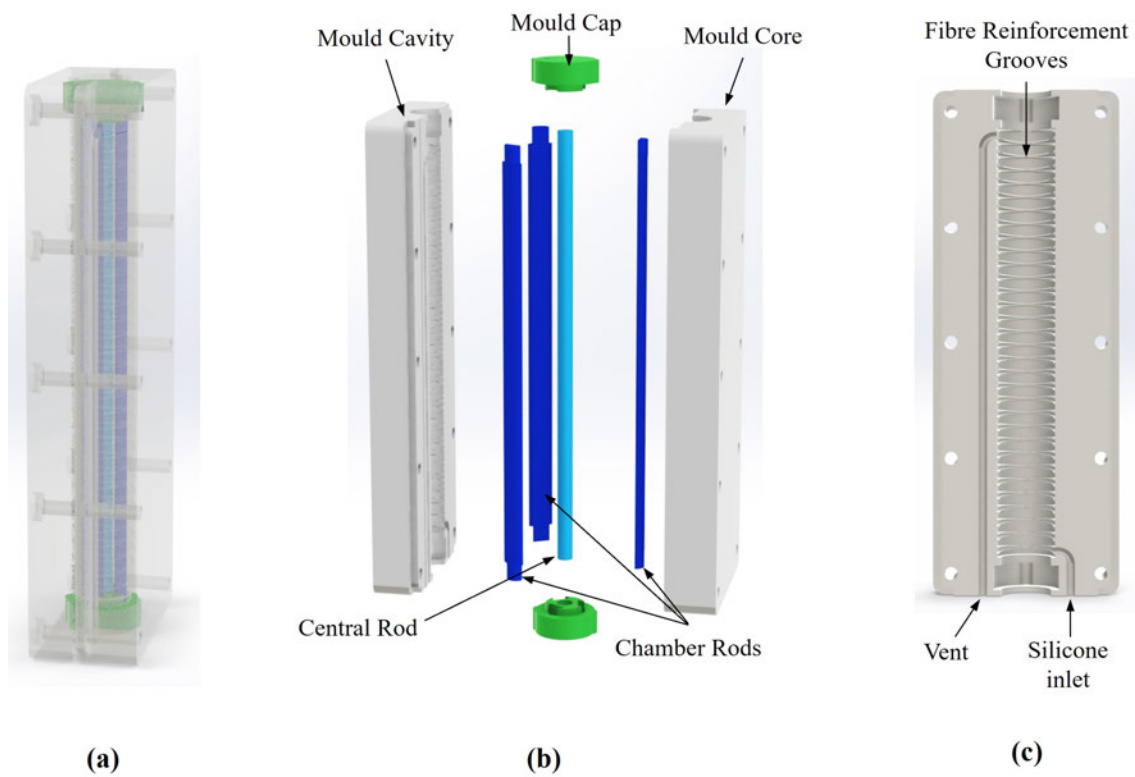


Figure 4.1: SCM mould design and its key components: (a) Assembled mould, (b) Exploded view of the mould, (c) Cross-sectional view of the main body.

4.2.1 Fabrication of the SCM

The active end-effector of the SCM is fabricated by injecting Elastosil M 4601 A/B silicone (Shore A hardness 28) into a 3D-printed mould. The schematic of the mould and its components, including the mould core and cavity, chamber and central rods, and mould caps, are illustrated in Fig. 4.1(b). The chamber rods generate the hollow hydraulic chambers and are fixed by the caps in such a manner that the chambers are positioned at 120°

intervals. The central rod forms the central channel, which can subsequently accommodate an endoscopic camera or additional instruments (see the SCM cross-section in Fig. 4.2(b) with further design parameters listed in Table 4.1).

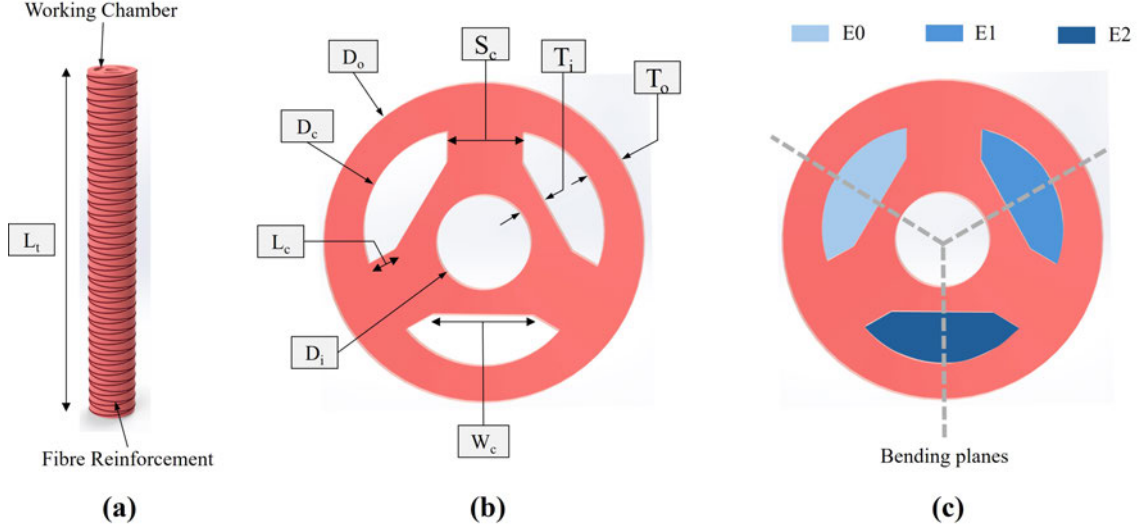


Figure 4.2: Fabricated SCM: (a) Overview of the length and fibre reinforcements, (b) Cross section dimensions, (c) Working chambers and their corresponding bending planes.

Table 4.1: Geometrical parameters of the 3D SCM design.

Parameter Description	Label	Value (mm)
Total length	L_t	106
Total outer diameter	D_o	14.6
Inner diameter	D_i	4.2
Chamber arc diameter	D_c	4.45
Chamber side length	L_c	1.33
Chamber width	W_c	4.5
Chamber spacing	S_c	3.38
Outer wall thickness	T_o	1.65
Inner wall thickness	T_i	1.15

As shown in Fig. 4.1(c), a double-wound helical groove with a pitch of 3.4 mm is incorporated into the mould core and cavity. This feature produces external channels intended for the secure placement of fibre reinforcement mitigating the radial expansion. The silicone casting process begins with the assembly and bolting of the mould core and cavity, along with the caps and internal rods, as depicted in Fig. 4.1(a). Parts A and B of Elastosil are mixed at a 9:1 ratio. Prior to injection, the mixture is placed in a vacuum chamber

to eliminate entrapped air bubbles. Following degassing, the mixture is injected into the mould via the designated inlet shown in Fig. 4.1(c), while maintaining the mould in the specified orientation. The silicone fills the mould from bottom to top until excess material exits through the vent, a strategy employed to minimise residual bubble formation.

The final **SCM**, shown in Fig. 4.2(a), has a length of 106 mm and a circular cross-section with a diameter of 14.6 mm. Given the inclusion of an endoscopic camera (more details in subsequent sections) with a central channel diameter of 4.2 mm, the outer diameter reaches 14.6 mm. This value may be reduced in future iterations to more closely match the dimensions of commercial colonoscopes (12.8 mm [225]). The length of the **SCM** was selected to fall within the typical range of the angulation (bending) section of standard colonoscopes employed in research [226].

The width of the flat surface of the working chamber, denoted by W_c in Fig. 4.2(b), is determined by the width of the **FPC** that is subsequently integrated onto this surface. Owing to the manufacturing constraints associated with the **FPC**, a value of 4.5 mm is selected for W_c to accommodate the substrate while providing sufficient tolerance for its insertion.

Given the cross-sectional geometry, the pressurisation of a single chamber induces bending of the **SCM** in the corresponding bending plane, which is oriented perpendicular to the flat surface. These bending directions are illustrated by dashed lines in Fig. 4.2(c). The colour-coded legend designates the naming convention for each chamber, which will be used in subsequent references.

4.2.2 FPC Design

The **SCM** developed in this work features a circular cross-sectional profile, differing from previously reported planar geometries. In such circular configurations, the neutral axis is centrally located. To facilitate positioning and integration, **FPCs** are mounted along the flat internal surface of the **SCM**'s working chamber, as performed in preceding sections. Although this location simplifies assembly, it resides at a radial distance from the neutral axis and, therefore, experiences measurable strain during actuation. This contrasts with earlier planar designs [14], where the flat surface coincided with the neutral axis under pure bending assumptions, resulting in negligible strain.

This radial offset introduces longitudinal strain in the **FPC**, expressed by Equation 4.1:

$$\varepsilon(y) = -\frac{y}{r} \quad (4.1)$$

where $\varepsilon(y)$ represents the strain at a radial distance y from the neutral axis, and r denotes the bending radius of the SCM. Referring to Fig. 4.2 and using the parameters listed in Table 4.1, the offset corresponding to the FPC plane under single-chamber pressurisation equals $y = -3.25$ mm, with a negative value, indicating below the neutral axis for the pressurised working chamber.

To ensure both mechanical compliance to the imposed strain and maintaining electrical integrity under deformation, the FPC was redesigned with a Kirigami-inspired architecture incorporating five integrated electrodes. This configuration draws upon principles of stretchable electronics outlined in [227]. The Kirigami pattern enhances flexibility and permits efficient routing of conductive traces within the constrained geometry of the working chamber.

The total number of electrodes was determined by three main constraints: (i) the 16-channel input limitation of the Quadra acquisition unit, (ii) the spatial allowance within the working chamber, and (iii) the resolution requirements for accurate electrical impedance measurements.

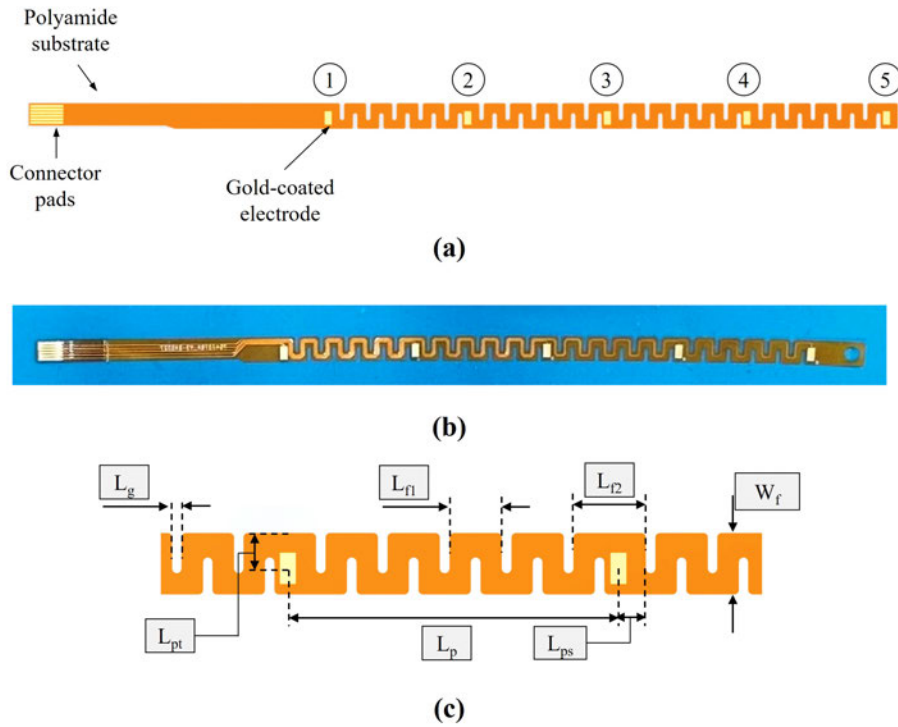


Figure 4.3: Designed FPC specifications: (a) Schematic of the design with the number of electrode pads, (b) The manufactured FPC, (c) Dimensions and design parameters.

The final schematic and manufactured design of the FPC is presented in Fig. 4.3(a,b), promising improved stretchability alongside reliable signal transmission and repeatable

Table 4.2: Pad-related geometrical parameters of the kirigami FPC.

Parameter Description	Label	Value (mm)
Distance between pads	L_p	21.3
FPC width	W_f	4.0
Non-pad segment length	L_{f1}	3.44
Pad segment length	L_{f2}	4.69
Pad side margin	L_{ps}	1.82
Pad top margin	L_{pt}	2.33
Gap length	L_g	0.6

performance. Fig. 4.3(c) highlights the critical dimensions of the stretchable layout, with numerical values summarised in Table 4.2. Several of these parameters were selected based on the fabrication constraints imposed by the supplier (JLCPCB, China), including a minimum trace width of 0.15 mm, trace-to-trace clearance of 0.09 mm, trace-to-pad clearance of 0.13 mm, and board outline clearance of 0.3 mm.

To interface each chamber's FPC with the Quadra device, a custom breakout board was developed. This board aggregates the output from individual FPCs and transmits the signals to the impedance spectroscopy. It incorporates 0.5 mm pitch, 6-pin connectors for the SCM-mounted FPCs, and a 16-pin connector for the flat extension cable connected to the Quadra system.

EIT protocol presented in this section reflects a revised configuration, incorporating changes in the number of electrodes embedded on the FPC, as compared to the previous chapters. The updated framework employs a five-channel protocol uniformly applied across all three working chambers, resulting in a total of 15 EIT channels. Given the results in Chapter 2, which demonstrated that good performance can be achieved with a smaller number of channels, this configuration enables comprehensive spatial coverage and supports multi-site impedance mapping. Functionally, channels 1 through 4 are designated to capture localised impedance variations, offering high spatial resolution within targeted regions. In contrast, channel 5 is designed to extend across a broader saline volume, thereby providing enhanced global sensitivity. Detailed injection (Ex_+ , Ex_-) and measurement (V_+ , V_-) electrode pairings are summarised in Table 4.3, corresponding to the electrode numbering denoted in Fig. 4.4 for reference.



Figure 4.4: FPC electrode numbering as a reference.

Table 4.3: EIT protocol used per chamber.

Channel	E_{x+}	E_{x-}	V_+	V_-
1	1	3	1	2
2	1	3	2	3
3	3	5	3	4
4	3	5	4	5
5	1	5	2	4

4.2.3 SCM Assembly

As illustrated in Fig. 4.5(a,b), three FPCs are inserted into the working chambers of the SCM and affixed to the flat internal surfaces using Sil-Poxy adhesive, applied to the rear of the polyimide substrate primarily beneath the electrode pads. While these attachment points remain fixed, the remaining segments of each FPC are left unconstrained to allow deformation within the SCM during pressurisation. Proper alignment of the FPC within each channel was essential to ensure uniform placement, which was one of the main reasons for adopting this integration approach.

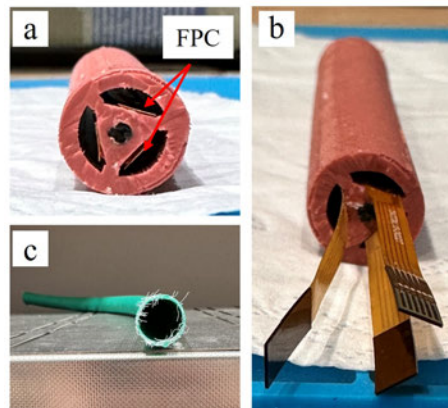


Figure 4.5: Assembling the SCC: (a) FPC integration inside working chambers,(b) FPC alignments relative to the proximal end, (c) Flexible overtube to facilitate insertion.

Subsequently, a customised endoscopic camera is introduced through the central channel to provide visual feedback during preclinical trials and to demonstrate the versatility of the system in accommodating additional surgical instruments. The camera (E-Vision Optoelectronic, China) features a head diameter of 3.9 mm, fibre optic illumination, and

USB video output. Its ultra-flexible wiring minimally impacts **SCM** compliance.

Following integration of both the camera and **FPCs**, the two ends of the **SCM** are sealed using Elastosil silicone caps to maintain overall softness. The proximal cap, designed to interface with hydraulic tubes, includes protruding silicone sleeves that envelop the tube ends. Upon insertion, the sleeves are secured using a constrictor knot, and an adhesive compound (Poly Kit, Tech Bond Solution) is applied to ensure complete sealing of the **SCM** inlets. The distal cap, fabricated with a central hole and a length of 15 mm, matching the camera head, accommodates the camera while ensuring a sealed termination.

4.2.4 Flexible Overtube

To facilitate insertion of the **SCC** during preclinical testing, a flexible overtube is attached to the proximal end of the **SCM**. As shown in Fig. 4.5(c), the overtube is 400 mm in length with an outer diameter of 17.5 mm, and comprises an aluminium spring coil encased in a fabric sheath. The connection between the overtube and the **SCM** is secured using a heat-shrink interface. In addition to assisting with handling, the overtube organises the hydraulic tubing and **FPC** extension cables exiting the **SCM**. Finally, the schematic and physical assembly of the complete embedded **SCC**, including the overtube, are presented in Fig. 4.6(a) and Fig. 4.6(b), respectively.

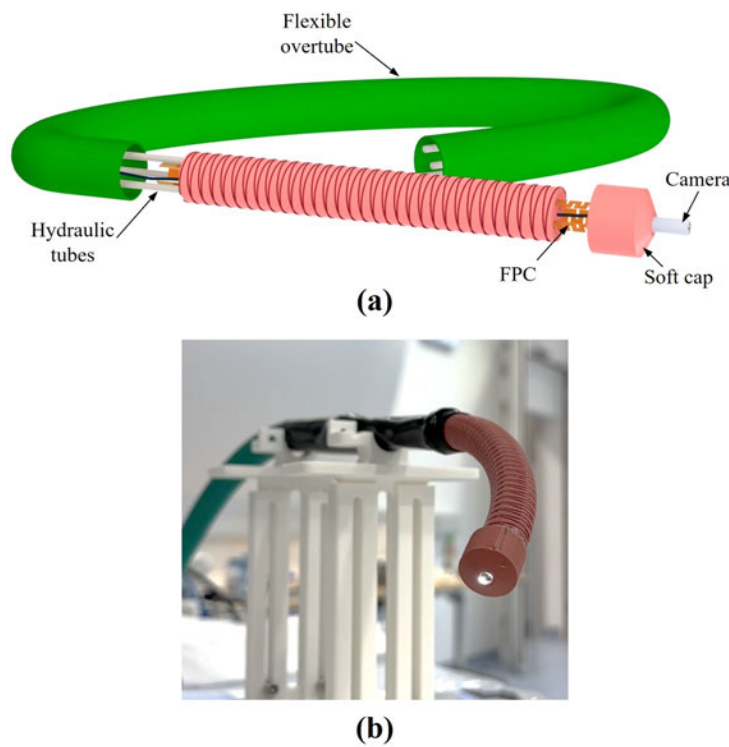


Figure 4.6: The complete **SCC**: (a) Schematics of the components, (b) Physical assembly.

4.3 MLP Architecture

To estimate the non-linear spatial shape of the **SCM** from **EIT**-derived impedance signals, an **MLP** network is employed. The **MLP** architecture is selected instead of the recurrent structures used previously, offering reduced computational complexity and seamless integration into closed-loop control feedback modules. Further justifications for this replacement are detailed in Section 4.7.

The input vector $\mathbf{I}$ comprises 18 features, formed by concatenating the min–max normalised values from 15 **EIT** channels with the hydraulic pressures from the three chambers:

$$\mathbf{I} = \begin{bmatrix} P_1 & P_2 & P_3 & R_1 & R_2 & \cdots & R_{15} \end{bmatrix} \quad (4.2)$$

The corresponding output vector $\mathbf{O}$ includes the **3D** coordinates of discrete key points located along the **SCM** backbone, expressed as:

$$\mathbf{O} = \begin{bmatrix} S_1 & S_2 & \cdots & S_m \end{bmatrix} \quad (4.3)$$

$$S_m = \begin{bmatrix} x_m & y_m & z_m \end{bmatrix} \quad (4.4)$$

where P_c denotes the pressure in chamber c , R_t the impedance from **EIT** channel t , and S_m the **3D** coordinates of the m^{th} key point. As shown in Fig. 4.7, the network architecture includes an input layer of 18 neurons, followed by ten fully connected hidden layers with 100 neurons each. ReLU is applied as the activation function across all hidden layers. The output layer is configured as a regression layer, with its size matched to the dimensionality of the target vector. Forward computations through the network follow:

$$\mathbf{a}^{(0)} = \mathbf{I} \quad (4.5)$$

$$\mathbf{g}^{(l)} = \mathbf{W}^{(l)} \mathbf{a}^{(l-1)} + \mathbf{b}^{(l)} \quad (4.6)$$

$$\mathbf{a}^{(l)} = \sigma^{(l)}(\mathbf{g}^{(l)}) \quad (4.7)$$

$\mathbf{W}^{(l)}$ and $\mathbf{b}^{(l)}$ represent the weight and bias matrices associated with the l^{th} layer. The ReLU activation function, denoted by $\sigma^{(l)}$, operates on the input $\mathbf{g}^{(l)}$, referred to as the pre-activation output. The resulting output of the activation function is given by $\mathbf{a}^{(l)}$, termed the post-activation output. The network is optimised using the Adam algorithm with an initial and constant learning rate of 0.001, a mini-batch size of 512, and a maximum of 30 training epochs. All hyperparameters were tuned through grid search. The

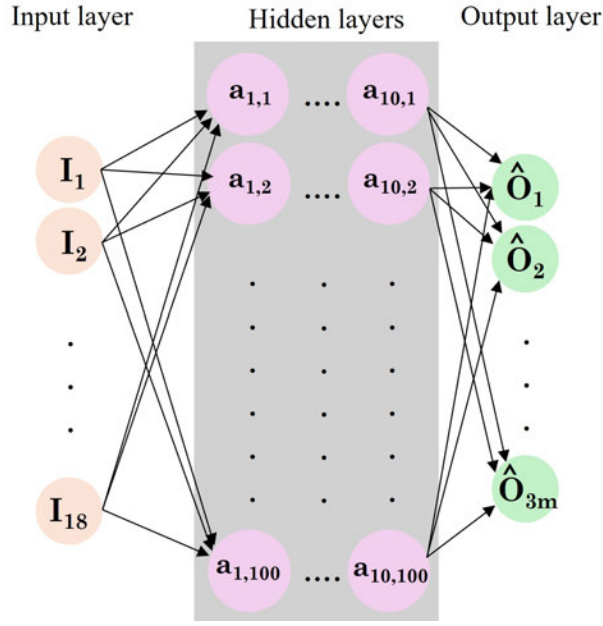


Figure 4.7: MLP architecture comprising input layer, hidden layers, and a regression layer. $a_{10,100}$ denotes the post-activation output of the 100th neuron of the 10th hidden layer.

loss function is defined as the Mean Squared Error (MSE):

$$\mathcal{L}_{\text{MSE}} = \frac{1}{n} \sum_{i=1}^n \sum_{j=1}^m (\hat{O}_{ij} - O_{ij})^2 \quad (4.8)$$

In this equation, $\hat{O}_{ij}$ and O_{ij} denote the predicted and ground truth values, respectively, where $i = 1, \dots, n$ indexes the training samples and $j = 1, \dots, m$ corresponds to the positional target data of marker j . To promote generalisability and prevent overfitting, L2 regularisation with $\lambda = 0.0001$ is included:

$$\mathcal{L}_{\text{total}} = \mathcal{L}_{\text{MSE}} + \mathcal{L}_{\text{reg}} \quad (4.9)$$

$$\mathcal{L}_{\text{reg}} = \lambda \sum_{l=1}^L \left\| \mathbf{w}^{(l)} \right\|^2 \quad (4.10)$$

The model is trained via backpropagation to minimise the total loss defined in Equation 4.9.

4.4 Experimental Setup

The experimental setup includes a control system and data acquisition units. The control system in Fig. 4.8(a) deforms the SCM by individually actuating each chamber using a

syringe pump system. Saline-filled syringes are driven by NEMA 17 stepper motors (1.8° resolution) via lead-screw transmissions, providing higher torque to accommodate the stiffer material of the **SCM** compared to previous designs. Unlike earlier configurations, three syringe pumps are dedicated to the chambers E0, E1, and E2. Pressurisation is performed in open-loop, with fluid delivered through chamber-specific tubing. Each motor is controlled via a DRV8825 driver (Texas Instruments, USA) mounted on a RAMPS 1.4 shield and interfaced with an Arduino Mega 2560.

Chamber pressures are monitored using MS5803-14BA sensors (TE Connectivity, Switzerland), sequentially addressed via a TCA9548A I²C multiplexer (Texas Instruments, USA) and transmitted to an Arduino Due. This unit interfaces with MATLAB 2022a through a serial connection for real-time data logging.

For **EIT** acquisition, electrodes embedded in the **FPCs** are linked to the Quadra system via flat flexible jumper cables (0.5 mm pitch) routed through the overtube to a breakout extension board. The system operates at a 0.05 ms sampling rate using a custom LabVIEW interface. **EIT** data are recorded as text files and post-processed offline to extract impedance values and timestamps. During experimental trials, the positional data of five

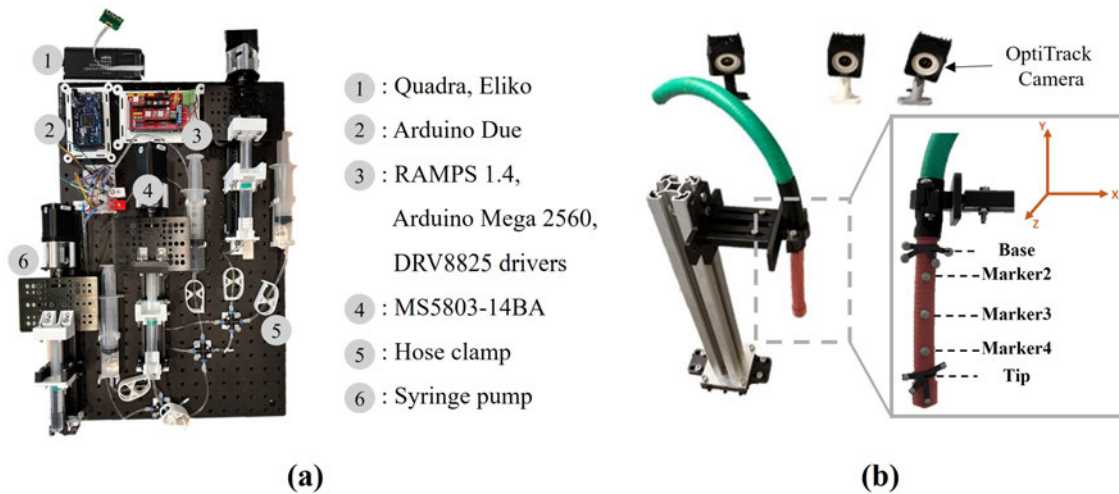


Figure 4.8: Experimental setup: (a) Control system and the list of its components, (b) Positional data acquisition setup.

key points along the **SCM** are captured using three OptiTrack cameras, enabling ground truth data collection. Intuitively, increasing the number of key points improves the resolution and accuracy of the reconstructed shape. However, the addition of optical markers introduces unintended external forces due to gravity. Furthermore, as discussed in the following chapter, the number of available **EM** sensors for tracking **SCM** positional data under *ex vivo* conditions is limited to three. Therefore, to minimise the discrepancy in the

number of tracked key points between the two domains (*ex vivo* and free motion), while still achieving higher spatial resolution than in *ex vivo* experiments, five key points were selected for tracking in the free motion studies.

The **SCM** is vertically anchored to maintain consistent visibility of all reflective markers throughout deformation. Fig. 4.8(b) illustrates rigid body markers at the base and tip, along with three individual markers evenly spaced in between. Rigid bodies enable the acquisition of both positional and orientational data via quaternion representation, allowing full shape reconstruction. Marker diameters are selected according to placement. Particularly, 6.4 mm markers for the backbone and 4 mm markers at the tip to reduce gravitational loading. Markers trajectories serve as ground truth for training the machine learning model by providing the **SCM** shape. This experimental setup and the fabricated **SCM** were employed in all subsequent experiments presented in this chapter.

4.5 Methods

4.5.1 FPC Deformation Analysis

To evaluate the reliability of **EIT** measurements under the newly proposed design, two experiments were conducted in parallel with integrating **FPC** into the **SCM**. The aim was to observe the out-of-plane deformation behaviour of the **FPC** under axial loading, both under in-plane and out-of-plane bending, and to assess the implications of such deformation on signal stability over extended durations.

As illustrated in Fig. 4.9, the central beam structure of the **SCM**, comprising the central instrument channel and the flat surfaces of the working chamber, was isolated for this study. A single **FPC** was affixed to one flat surface. The assembled unit was mounted on two custom-designed 3D-printed fixtures, as shown in Fig. 4.10, to simulate in-plane and out-of-plane bending of the central beam. The in-plane fixture, designed to stretch the **FPC**, forms a 90° arc with a bending radius of 57.8 mm (Fig. 4.10(a)). The out-of-plane fixture compresses the **FPC** over a 120° arc with a 56.2 mm bending radius, simulating deformation caused by in-plane bending of an adjacent chamber oriented 120° from the passive one (see Fig. 4.1(c)). The larger bending angle was selected to emphasise out-of-plane deformations.

The entire assembly was subsequently submerged in a 1% saline solution (as in Fig. 4.11) and maintained under static conditions for a duration exceeding 3000 s, while **EIT** data were acquired at 61 kHz spectrogram frequency from the first four channels in Table 4.3. The corresponding outcomes are discussed in the Results section.

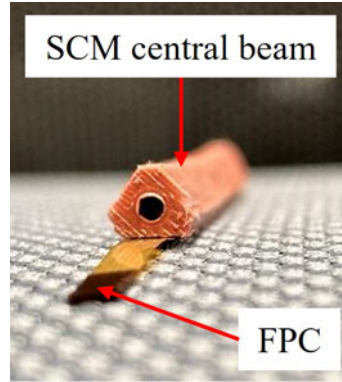


Figure 4.9: FPC central beam to simulate out-of-plane deformation.

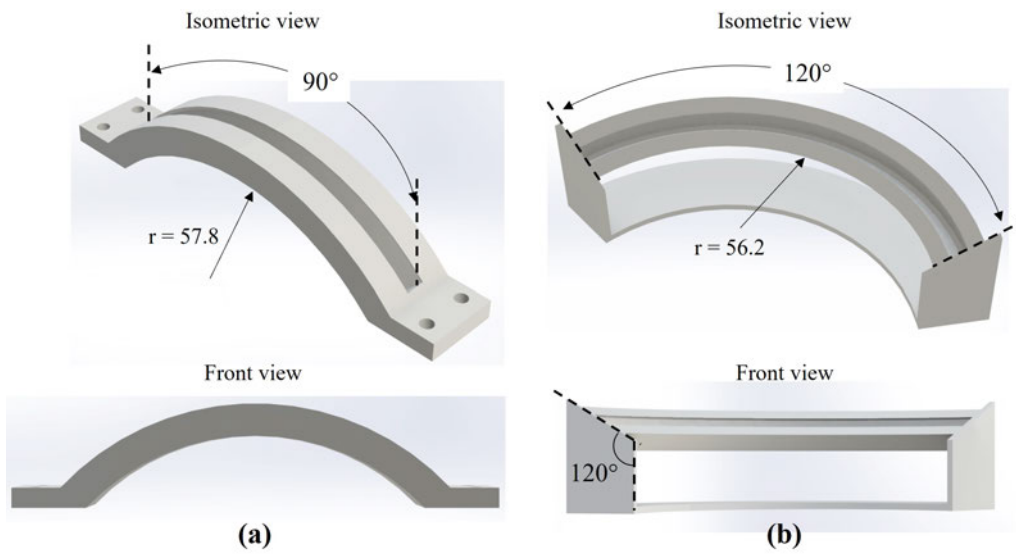


Figure 4.10: Fixtures to introduce tension and compression to the FPC: (a) In-plane bending (b) Out-of-plane bending fixture.

4.5.2 EIT Characterisation Under the New FPC Design

To assess the performance of EIT under dynamic and quasi-static actuation, and to account for the influence of SCM design and working chamber configuration, a series of experiments was conducted using the 3D SCM.

Signal Repeatability and Robustness

Chambers E0, E1, and E2 were independently actuated using quasi-static triangular waveforms at an average frequency of 0.04 Hz (25.3 s per cycle). Each chamber underwent ten full inflation-deflation cycles while the adjacent chambers remained unpressurised, inducing in-plane bending in three orthogonal planes. The actuation pressure amplitude was fixed at 2.3 bar.

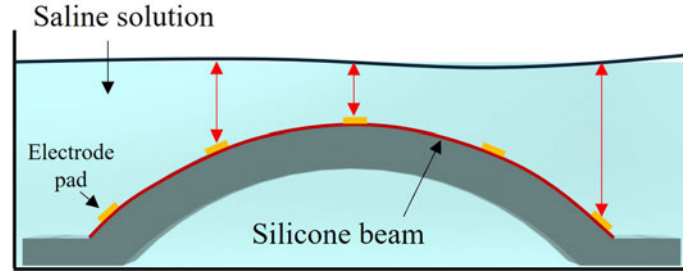


Figure 4.11: Schematic of submerged fixture in saline. Red arrows denote varying saline depth above each electrode pad.

The influence of individual chamber pressurisation on adjacent chambers and the corresponding EIT responses is analysed, followed by examining the pressure–EIT signal relationship. To quantify signal robustness, the Average Standard Deviation (ASD) across the five EIT channels was computed. First, the EIT and pressure data were segmented by cycle and interpolated to uniform length. For each EIT channel ($ch = 1:5$), standard deviation (SD_k) values were calculated at each of the K pressure points within a cycle, and the ASD was determined as:

$$ASD_{ch} = \sqrt{\frac{SD_1^2 + SD_2^2 + \dots + SD_k^2}{K}} \quad (4.11)$$

Additionally, to examine EIT sensitivity, the EIT signal variation was plotted as a function of tip displacement using positional tracking data. This analysis reveals the consistency of EIT responses across different deformation magnitudes.

Hysteresis Analysis

To assess whether the stretchability of the redesigned FPC contributes to energy dissipation in the mechanical response of the SCM, hysteresis experiments were conducted. Each chamber was actuated using sinusoidal pressure inputs with amplitudes ranging from 0.9 to 2.25 bar at an average frequency of 0.09 Hz (10.92 s per cycle). A higher actuation rate was selected compared to the previous tests, as hysteresis effects become more pronounced at elevated frequencies due to increased energy dissipation [228]. In total, 36 cycles of pressure and corresponding EIT data were recorded from each chamber. Hysteresis was quantified for each chamber using Equation 4.12, where H denotes the hysteresis percentage, $\Delta R_{\max}$ is the maximum impedance discrepancy between inflation and deflation phases, and $R_{\max} - R_{\min}$ represents the impedance range:

$$H = \frac{\Delta R_{\max}}{R_{\max} - R_{\min}} \times 100 \quad (4.12)$$

The analysis includes a comparison between channels 3 and 5 to investigate the effect of increasing electrode separation on hysteresis. Additionally, to separate the hysteresis of the actuator, due to its intrinsic viscoelasticity, and the hysteresis of the sensing, the inflation-deflation loop of tip displacement versus pressure was analysed. Tip displacement was computed as the Euclidean norm of its Cartesian coordinates:

$$d = \sqrt{x_{\text{tip}}^2 + y_{\text{tip}}^2 + z_{\text{tip}}^2} \quad (4.13)$$

FPC Impact on Flexibility

Despite the added stretchability introduced by the kirigami design, the material mismatch between the polyamide-based FPC (Young's modulus of 1600 MPa) and the silicone body (Young's modulus of 6.3 MPa) necessitates evaluation of the FPC's effect on the flexibility of the continuum manipulator. To this end, an SCM without embedded FPCs was actuated under identical quasi-static pressurisation protocols as described in Section 4.5.2. The resulting tip trajectories, both in individual Cartesian components and in terms of total displacement, were computed and compared against those of the FPC-integrated SCM. The displacement data were plotted against pressure values to assess potential reductions in deformation capacity attributable to the presence of the FPC. Comparative results are presented in the Results section.

4.5.3 MLP Training for Shape Prediction

Data were collected under both single- and multi-chamber pressurisation protocols. In single-chamber trials, each chamber was actuated individually using randomly selected pressure values within a defined range to promote generalisability of the data-driven model. Each session lasted approximately 30–40 minutes, yielding a total of 85,245 motion samples. Multi-chamber trials involved the sequential random activation of two chambers simultaneously, with randomised timing, order, and pressure magnitudes. These conditions produced 65,537 motion samples and allowed the SCM to bend across a continuous range of planes, in contrast to the more limited trajectories observed under single-chamber actuation. The data acquisition and learning pipeline for free motion shape reconstruction is illustrated in Fig. 4.12.

EIT, pressure sensors, and OptiTrack sampled datasets at average actual rates of 13.2 Hz, 11.1 Hz, and 114.5 Hz, respectively. The multi-modal sampled datasets were synchronised using timestamp alignment. Due to differences in sampling rates, PCHIP interpolation was applied to achieve temporal synchronisation. EIT signals were baseline-corrected using pre-pressurisation measurements, and all positional data were referenced

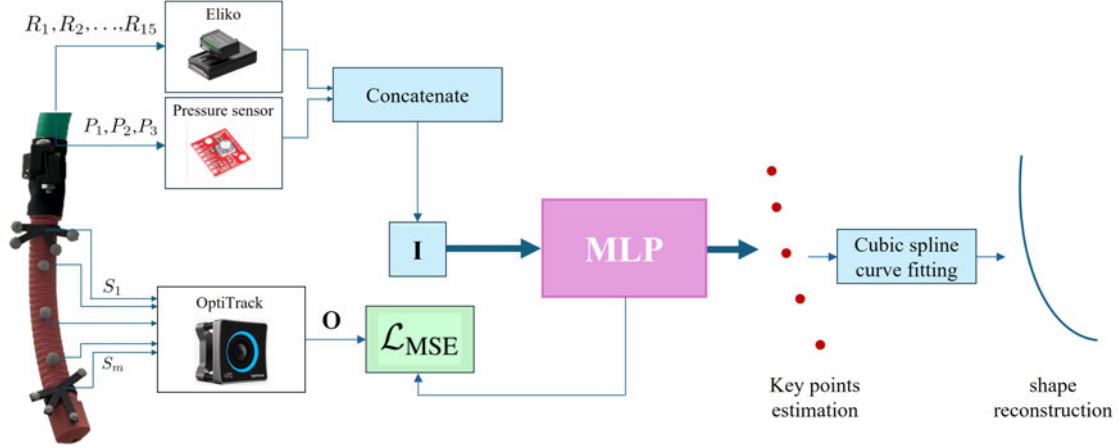


Figure 4.12: Free motion shape reconstruction pipeline: data collection, post-processing, training, and curve fitting.

to the base marker. The dataset was partitioned into training, validation, and test sets using a 70:15:15 ratio. For shape reconstruction, cubic spline fitting was employed to generate continuous curves from discrete key-point positions in both ground-truth and predicted data.

In terms of the predominant curve-fitting approaches, B'ezier curves are among the most widely used [5, 74, 229]. However, in this chapter, the choice of cubic splines over B'ezier curves is due to their greater flexibility. B'ezier curves are defined by a single global formulation, which limits their local adaptability, whereas splines allow interpolation between control points, enabling localised shape adjustments [230]. It remains challenging to reliably define soft robotics kinematics by a specific mathematical formulation, particularly under complex and uncertain environmental conditions [231]. Consequently, B'ezier or PCC representations may become invalid in such scenarios. Nonetheless, cubic splines provide a spatially continuous representation of the shape while offering additional flexibility in modelling the geometry of the SCM.

4.6 Results

4.6.1 FPC Deformation Analysis

Out-of-plane deformation of the FPC under tension, when embedded within the active chamber, is observed to be negligible for bending angles up to 90° and within the designated bending radius, as evidenced by the top and side views in Fig. 4.13(a). It is assumed that during tensile loading, the rear surface of the FPC maintains full contact with the underlying silicone beam, thereby restricting any out-of-plane deformation normal to the

beam surface due to this continuous adherence.

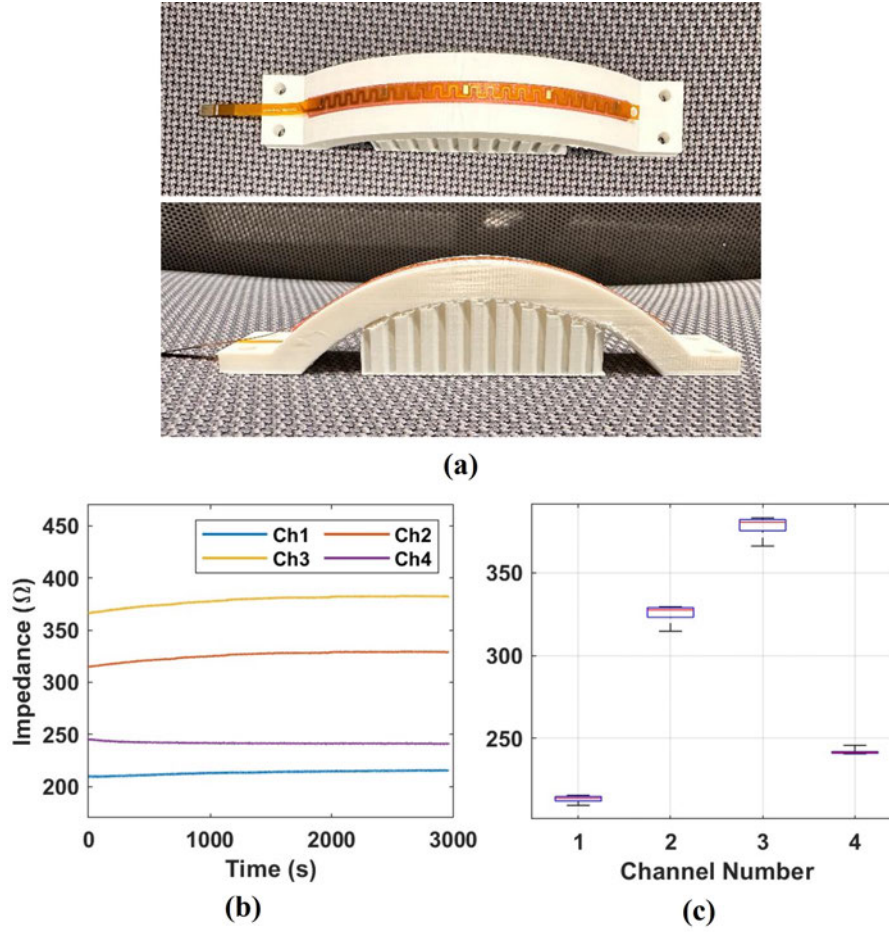


Figure 4.13: Analysing kirigami FPC under static tension: (a) Out-of-plane deformations observations, (b) Signal temporal stability in four channels, (c) Box plot of stability analysis.

The EIT signal stability over a 3000 s period, shown in Fig. 4.13(b), indicates consistent output while the FPC is subjected to longitudinal stretching. Ideally, if the FPC were mounted on a flat central beam, all four channels would exhibit similar impedance values, given that the electrode pads are designed to interrogate equivalent volumes of saline. However, in the current curved configuration, non-uniform impedance values are recorded across the channels. This discrepancy is attributed to two primary factors: (i) the stretching of the FPC alters the inter-electrode distances unevenly along its length, thereby affecting the effective sensing volume for each channel; and (ii) the curvature of the assembly within the saline container results in variable fluid depth above each channel (see Fig. 4.11). Channels 1 and 4, located at the ends of the FPC arc and immersed more deeply, register lower impedance due to greater saline volume, whereas channels 2 and 3, situated at shallower depths near the arc centre, exhibit higher impedance. Fig. 4.13(c)

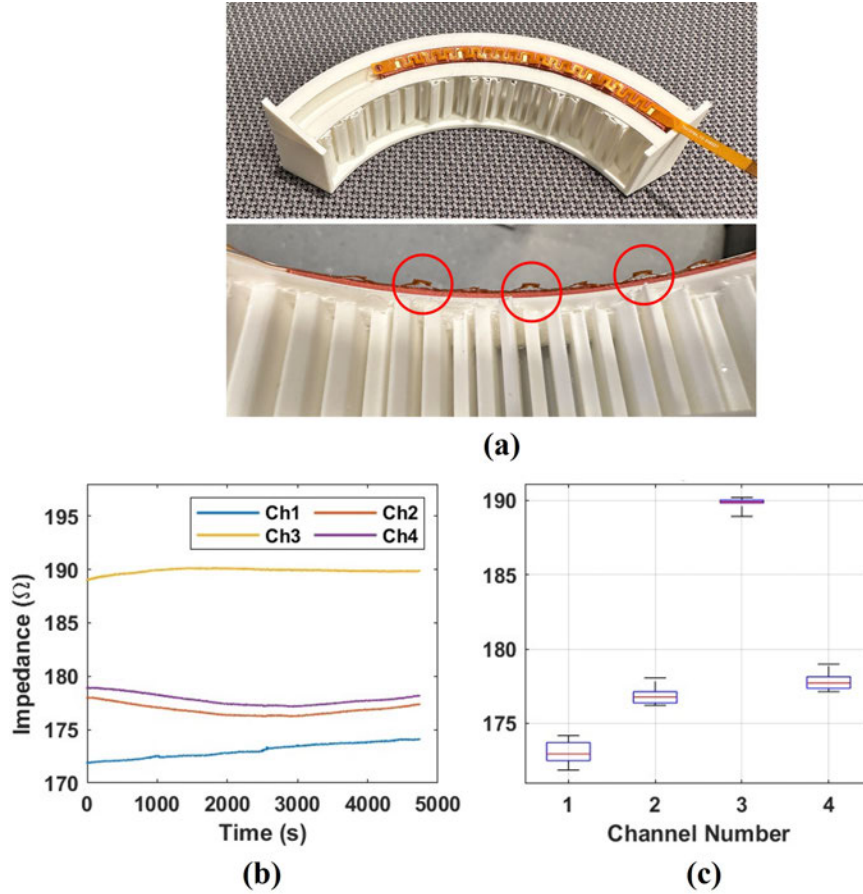


Figure 4.14: Analysing kirigami FPC under static compression: (a) Out-of-plane deformations observations with red circles denoting detached regions, (b) Signal temporal stability in four channels, (c) Box plot of stability analysis.

presents the temporal stability of impedance signals per channel. Channel 3 shows the highest variation, increasing from 366Ω to 382Ω ($\Delta 16 \Omega$), while channel 4 exhibits the lowest fluctuation, with a 5Ω range.

Under compressive loading, however, out-of-plane deformations are more prominent, particularly in regions not bonded to the silicone substrate by Sil-Poxy adhesive, as highlighted in Fig. 4.14(a). In contrast to the tensile condition, these deformations are unconstrained, leading to greater deviation of the FPC from the beam surface. The resulting partial detachment likely contributes to the channel variance observed in Fig. 4.14(b) as the separation interferes with the current flow. Additionally, variations in saline depth across the curved geometry, as in the tension case, further affect the measured impedance.

Over 4700 s, channels 1, 2, and 4 maintain impedance values within $171.9\text{--}178.9 \Omega$, while channel 3, located centrally along the arc, reports a median of 189.9Ω . Stability analysis in Fig. 4.14(c) shows that channel 1 experiences the highest variation of 2.4Ω

(1.3% of its median), whereas channel 4 has the lowest at 1.3Ω (0.7%).

Given the stability of the signals produced by the kirigami FPC under both tension and compression over an extended period, and based on the results determining that deformations of the FPC within the working chamber cause only minor variations in the EIT readings, the current design can be regarded as reliable for use in the subsequent experiments.

4.6.2 EIT Characterisation Under the New FPC Design

Signal Repeatability and Robustness

The triangular pressure signals in Fig. 4.15(a) demonstrate the activation of the working chamber E1, with pressure values ranging from 936 to 2318 mbar. Although E0 and E2 remain passive, their pressure values increase in the same trend as the active chamber, though with a reduced range, from 1001 to 1375 mbar for E0 and from 920 to 1247 mbar for E2. This increase is attributed to mechanical coupling between the chambers. As the active chamber inflates and bends, the other two chambers undergo compression, resulting in a reduction of the cross-sectional area of their respective working chambers. The similar pressure ranges observed in the passive chambers (a 47 mbar difference due to fabrication inaccuracies) are primarily due to the symmetric positioning of the passive chambers with respect to the bending plane of the active chamber.

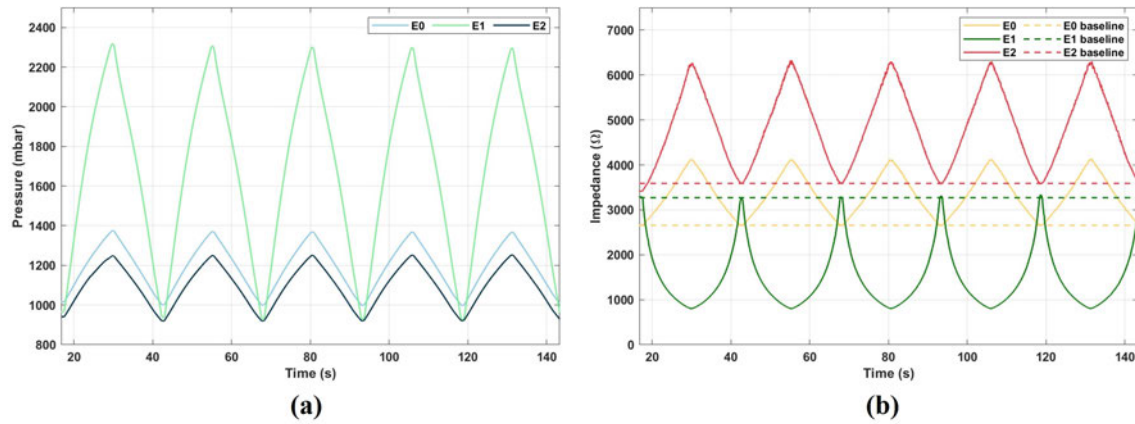


Figure 4.15: Evaluating EIT signal stability and repeatability. (a) Chambers pressure with E1 being active, (b) Corresponding EIT outputs of channel 5.

In Fig. 4.15(b), the EIT values of channel 5, corresponding to the active chamber (E1), decrease as the pressure in E1 increases (Fig. 4.15(a)). This correlation between pressure and impedance was previously identified in Chapter 2 for a single working chamber. In the multi-chamber configuration, however, the EIT readings of the passive chambers (E0

and E2) are also influenced by the pressure increase in E1. Specifically, the **EIT** values of E0 and E2 rise as their respective pressures increase. This behaviour results from the reduction in the cross-sectional area of E0 and E2 caused by the expansion of E1, leading to a corresponding increase in **EIT** values. The impedance change in E1 is 2465 Ω , whereas it is 1472 Ω and 2736 Ω in E0 and E2, respectively.

Furthermore, the **EIT** variations in the passive chambers (E0 and E2) exhibit a relatively linear trend, unlike those in the active chamber (E1). This linearity indicates that impedance changes in E0 and E2 are primarily driven by electrode spacing, consistent with Pouillet's law, where impedance varies linearly with length. In contrast, the **EIT** variations in E1 arise from concurrent changes in both cross-sectional area and length caused by chamber expansion under pressurisation.

Although the pressure increase in E2 is only 27% of that in E1, the impedance variation in E2 exceeds that of E1. This outcome is attributed to the dependency of the **EIT** signal on the cross-sectional volume of the conductive solution. Given the non-automated nature of the manufacturing process and concerns regarding its repeatability, the chambers of the **SCM** are unlikely to possess identical geometries. Specifically, the cross section of E2 appears smaller than that of E1, so further compression yields a greater impedance change compared to the impedance change caused by the expansion of E1. This geometric discrepancy is also evident in the horizontal dashed lines representing baseline impedance values in Fig. 4.15(b), which illustrate that even in the unpressurised state, the **EIT** values differ across chambers, despite identical solution concentration and fabrication protocol. The baseline values are 2655 Ω , 3271 Ω , and 3590 Ω for E0, E1, and E2, respectively. Based on the baseline of E2, it is inferred that its initial cross section is the smallest among the three. Additionally, the rate of impedance change in E1 differs from that in the passive chambers, with a decreasing slope, indicated by the green line. This behaviour is related to the hyperelastic properties of silicone, discussed later in the section.

To better understand the correlation between **EIT** channels and chamber pressure, Fig. 4.16 presents the mean impedance of all five channels per chamber, plotted against the pressure in the active chamber (E1). The **EIT** values are normalised with respect to their baseline. In the passive chambers E0 and E2, impedance increases with pressure, while in E1 it decreases. Across all chambers, channels 1 to 5 exhibit similar impedance trends, with channels 5 showing the highest values. Recall that channel 5 employs electrodes with greater separation than channels 1 to 4, resulting in larger impedance variations in both active and passive chambers. However, channel 4 exhibits unexpectedly divergent behaviour despite having the same electrode spacing as channels 1-3. At 2149 mbar, channel 4 exhibits lower impedance by 278.6 Ω in E0 and 385.8 Ω in E1, compared to

the mean of channels 1 to 3. This anomaly may be attributed to non-uniform chamber deformation, potentially caused by uneven fibre reinforcement tightness, which could result in greater expansion near channel 4. Additionally, variations in electrode contact impedance may contribute. In E1, a sudden impedance drop is observed in channel 4 at 1167 mbar, attributed to the presence of a bubble in the conductive solution affecting only this channel, as no such change is noted in the others.

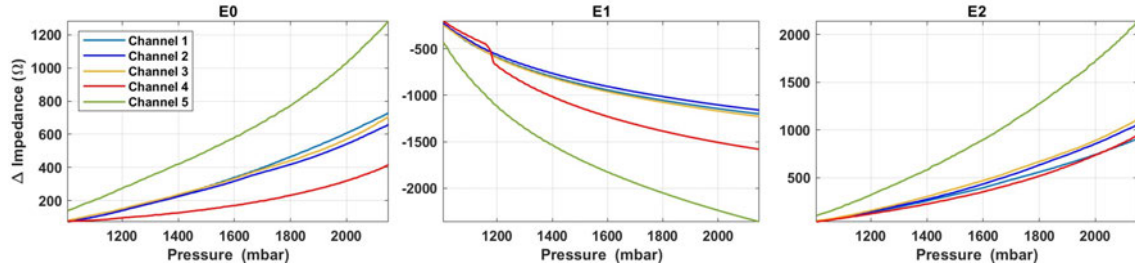


Figure 4.16: Relationship between EIT channels output and pressure in passive and active chambers. E1 is active here.

Signal Standard Deviation

Repeatability of the signal was evaluated by calculating the ASD of each channel over ten full inflation-deflation cycles in Fig. 4.17. Channel 5 again exhibits the highest variation, particularly in E1 while active, with a maximum ASD of 21.6 Ω , representing 0.9% of the highest EIT value recorded for E1 channel 5 in Fig. 4.16 (2359 Ω).

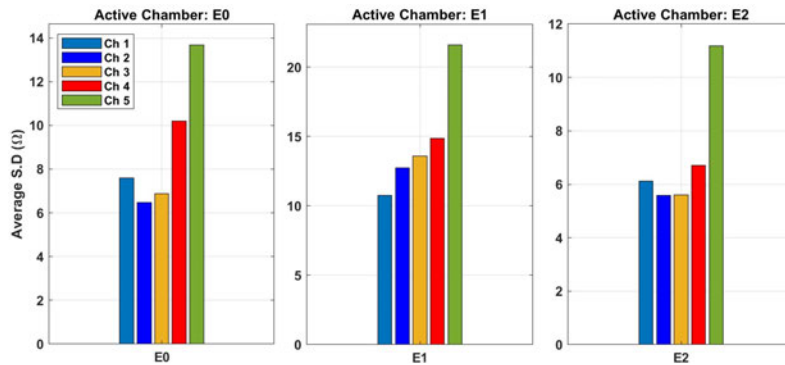


Figure 4.17: ASD of EIT channels in their corresponding active chamber.

EIT Sensitivity to Tip Displacement

Fig. 4.19 shows that the absolute rate of impedance change decreases with increased displacement (see Fig. 4.18), indicating reduced sensitivity by presenting the mean EIT value of channel 5 in E0 plotted against tip displacement. Fig. 4.19(b) shows the corresponding derivative, highlighting a steep initial slope, which represents the EIT dead

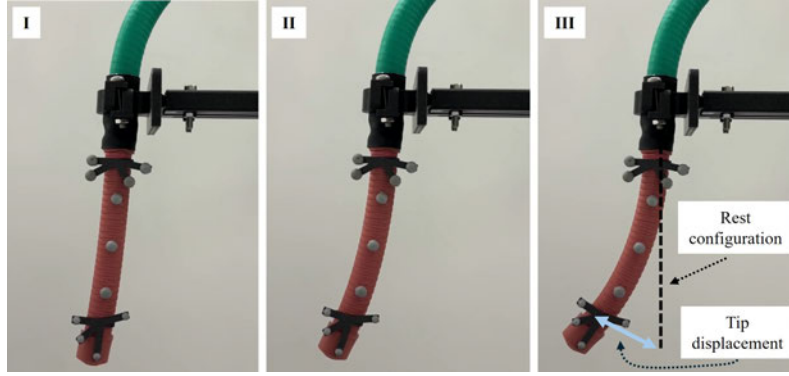


Figure 4.18: Increasing tip displacement as the SCM bends in plane.

zone. In this region, pressure increases result in reduced impedance (due to chamber expansion), but no significant tip displacement is observed, since the pressure required to deflect the entire SCM is greater than that needed for internal chamber expansion. Consequently, the derivative initially assumes large values. As shown in the magnified view in Fig. 4.19(c), the absolute derivative declines and approaches zero, indicating that further expansion becomes less significant. The derivative decreases from $300 \text{ } \Omega/\text{mm}$ at 2 mm displacement to $10.5 \text{ } \Omega/\text{mm}$ at 36.5 mm displacement. This trend aligns with the elastic behaviour of silicone rubber, as determined in its stress-strain response [232]. According to models such as Neo-Hookean, higher stress levels are required for equivalent strain at large deformations. Therefore, as input pressure increases, the resulting strain, and hence chamber expansion, progresses at a slower rate. As EIT reflects changes in cross section, the impedance change rate similarly decreases.

Hysteresis Analysis

The input pressure and the corresponding EIT values in Fig. 4.20(a,b), with E1 being active, have a higher frequency compared to those in Fig. 4.15. This experimental condition is designed to evaluate the hysteresis of the measurement method. The same procedure is applied to E0 and E2 by repeating the actuation to assess hysteresis in those chambers as well. Table 4.4 provides an overview of the hysteresis, calculated using Equation 4.12, for different active chambers and EIT channel numbers. The variation in hysteresis observed across chambers for the same channel number may be attributed to fabrication-related factors, such as differences in cross-sectional dimensions, non-uniformity in the silicone mixture, and inconsistent distribution of microbubbles within the silicone matrix. Additionally, the presence of air bubbles in the saline solution cannot be entirely ruled out, despite an intensive de-bubbling phase carried out prior to testing.

The results also show that, within the same chamber, hysteresis decreases as the effective

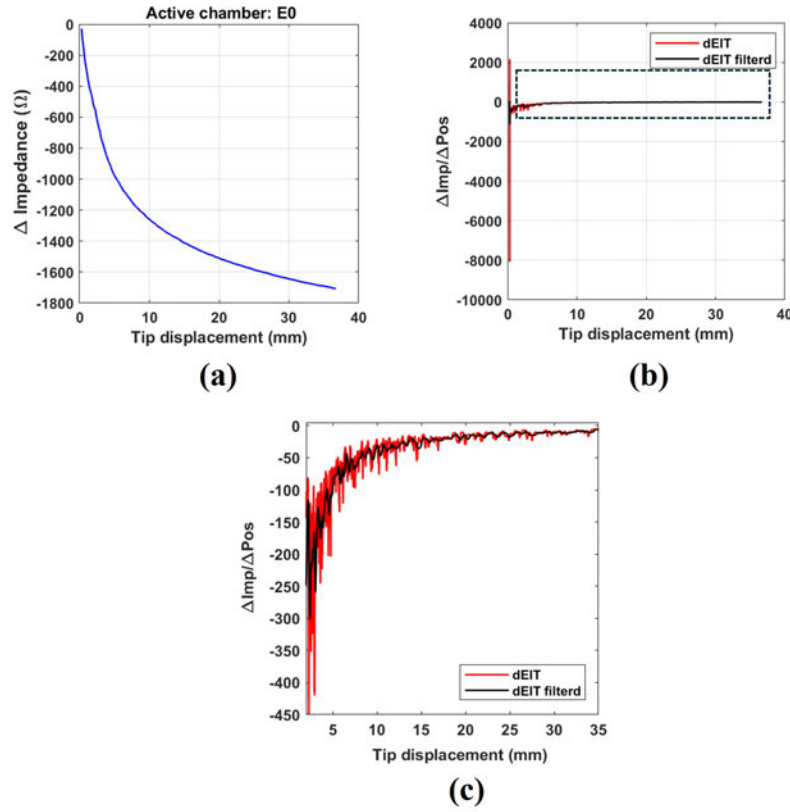


Figure 4.19: EIT sensitivity to tip displacement assessment: (a) Relationship between the baseline-corrected value of channel 5 in active chamber E0 and tip displacement, (b) First derivative of impedance with respect to tip displacement, (c) Magnified view of the selected portion of derivative in (b).

Table 4.4: EIT hysteresis across chambers for two different channels represented by their corresponding numbers and diagrams. Red and blue marks denote current injection and voltage measurement pairs, respectively.

Channel	E0	E1	E2
 3	13%	18.2%	21%
 5	11.7%	17.3%	20%

measured volume increases, moving from channel 3 to channel 5. This trend is primarily due to the reduced value of $R_{\max} - R_{\min}$ in channel 3, which leads to proportionally higher hysteresis values according to the definition in Equation 4.12.

The hysteresis cycles illustrated in Fig. 4.21 clearly demonstrate that the range of impedance changes in channel 3 is lower compared to channel 5. It can also be observed that, across different chambers, the EIT values in chamber E2 exhibit greater robustness and repeatability, likely due to more stable FPC adhesion in the chamber, while E1 shows the least

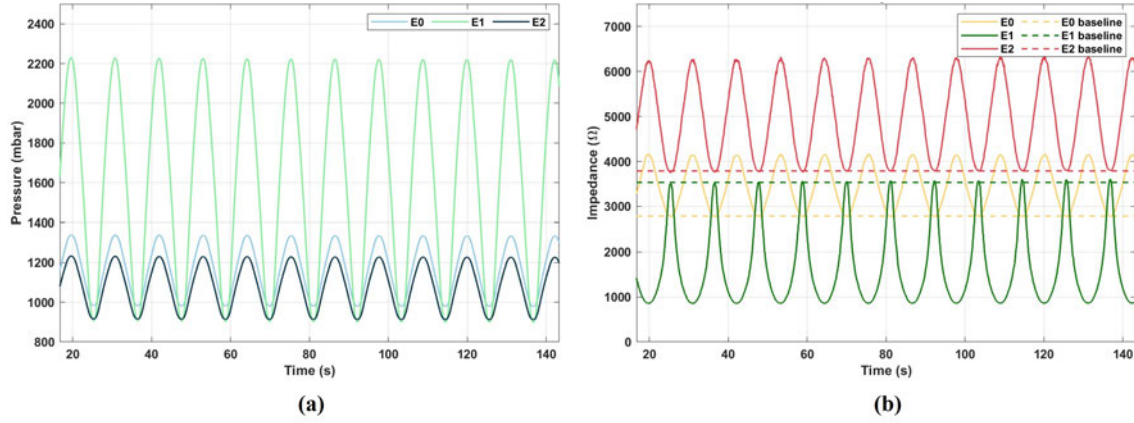


Figure 4.20: EIT hysteresis analysis: (a) Input pressure value with E1 being active, (b) Corresponding EIT outputs of channel 5.

repeatability, which aligns with the findings presented in Fig. 4.17. Additionally, in all cases shown, the impedance values are consistently lower during the deflation phase compared to the inflation phase.

This behaviour can be explained by considering the hyperelastic nature of the silicone material. During inflation, the working chambers of the SCM expand. However, in the subsequent deflation, the reduction in pressure does not immediately result in a corresponding contraction. Owing to the delayed response caused by the nonlinear elasticity of the silicone, the cross-sectional area remains larger than during inflation at the same pressure. Fig. 4.22 supports this explanation by illustrating that the inherent hysteresis of the silicone rubber significantly contributes to the overall hysteresis observed in the EIT measurements. The figure shows a delayed contraction of the SCM during deflation, indicating that a similar delay occurs in the volume of the working chamber, which consequently induces hysteresis in the impedance values.

FPC Impact on Flexibility

The single chamber pressurisation of E0, E1, and E2 results in in-plane bending, with the corresponding tip trajectories shown in Fig. 4.23(a) along each axis, as well as their total tip displacement plotted against the applied pressure. While the tip displacement is computed according to Equation 4.13 and is thus always positive, the x , y , and z components reflect the motion of the SCM relative to the global coordinate frame defined in Fig. 4.8(b). The displacement divergence shown in Fig. 4.23(b) represents the difference between the SCM tip motion with and without the FPC. These errors are computed as $d_{\text{fpc}} - d$ for total displacement and in a similar fashion for each axis. It is evident that, as pressure increases, the trajectories of the SCM with and without the FPC diverge pro-

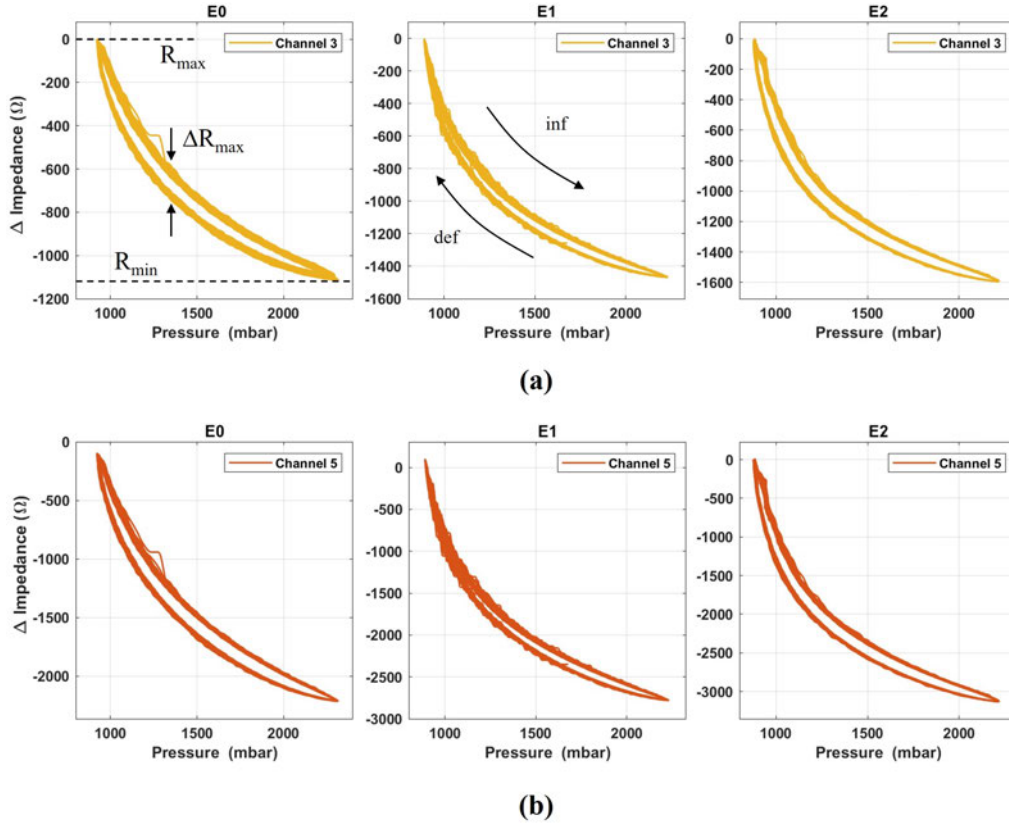


Figure 4.21: EIT hysteresis results in active chambers: (a) Channel 3, (b) Channel 5. Inflation and deflation cycles are indicated by arrows in (a) E1.

gressively in Fig. 4.23(a), and the corresponding errors in Fig. 4.23(b) show an increasing trend in the absolute error values. The maximum absolute displacement errors observed at 2000 mbar pressurisation are 4.7 mm, 1.4 mm, and 4.5 mm for chambers E0, E1, and E2, respectively. In chamber E1, the tip displacement error is negative, whereas it is positive for E0 and E2. A positive error indicates that the SCM with the FPC achieves a greater tip displacement compared to the one without the FPC, whereas a negative value indicates the opposite. The negative error observed in E1 is expected due to the stiffer nature of polyamide relative to Elastosil. The positive errors in E0 and E2 can be justified by the fact that the FPC is affixed to the flat surface of the SCM and it may locally restrict the longitudinal stretch of the chamber despite the design improvements aimed at enhancing flexibility. As a result, in chambers E0 and E2, reduced axial stretch may redirect the input pressure to promote bending rather than elongation, thereby increasing tip displacement.

This effect can be further substantiated by examining the error component along the y-axis, which corresponds to axial stretching. Across all chambers E0, E1, and E2, the error in y is consistently negative, specifically at lower pressure ranges, suggesting reduced elongation in the SCM with the FPC compared to the one without it. However, the

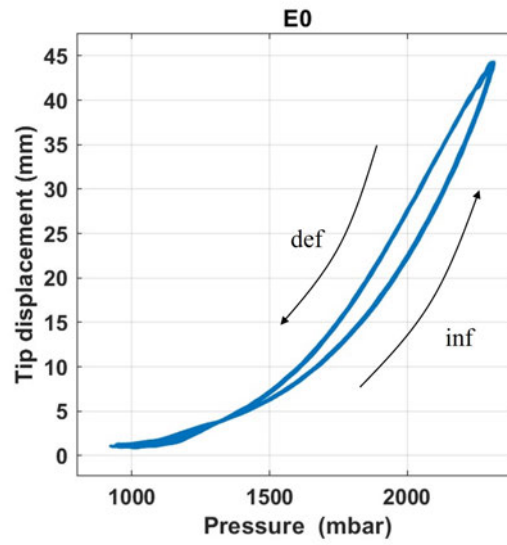


Figure 4.22: Intrinsic hysteresis of silicone rubber observed in tip displacement vs pressure.

observed errors may be influenced by fabrication inconsistencies. The **FPC** is manually affixed to the working chambers, and its positioning is inherently variable, potentially affecting the bending response of the **SCM**.

The results of this section demonstrate that integrating the **FPC** does not significantly restrict the motion of the **SCM**. Although the tip displacement with the **FPC** is slightly reduced compared to the configuration without it, the same workspace can still be achieved by applying a marginally higher pressure.

4.6.3 Ground Truth Positional Data Acquisition

In Fig. 4.24(a), the motion instances of the **SCM** under multi-chamber pressurisation span approximately 20 seconds, during which all three chambers are actuated sequentially with randomised order, timing, and pressure magnitude. The resultant positional data, recorded via OptiTrack from selected keypoints, are illustrated in three dimensions and along individual coordinate axes in Fig. 4.24(b). The trajectories reflect the actual configuration of the **SCM** during data collection and align with the coordinate frame shown in Fig. 4.8(b), with the base marker located at the top. The workspace depicted in Fig. 4.24(b) represents a subset of the **SCM**'s full operational volume, with the tip reaching a maximum Euclidean displacement of 54.74 mm. The maximum displacements of markers 4, 3, and 2 are 34.78, 15.49, and 3.92 mm, respectively. These values correspond to 63.5%, 28.3%, and 7.2% of the tip displacement, respectively.

Analysis across individual coordinate axes reveals that deflections in the x and z direc-

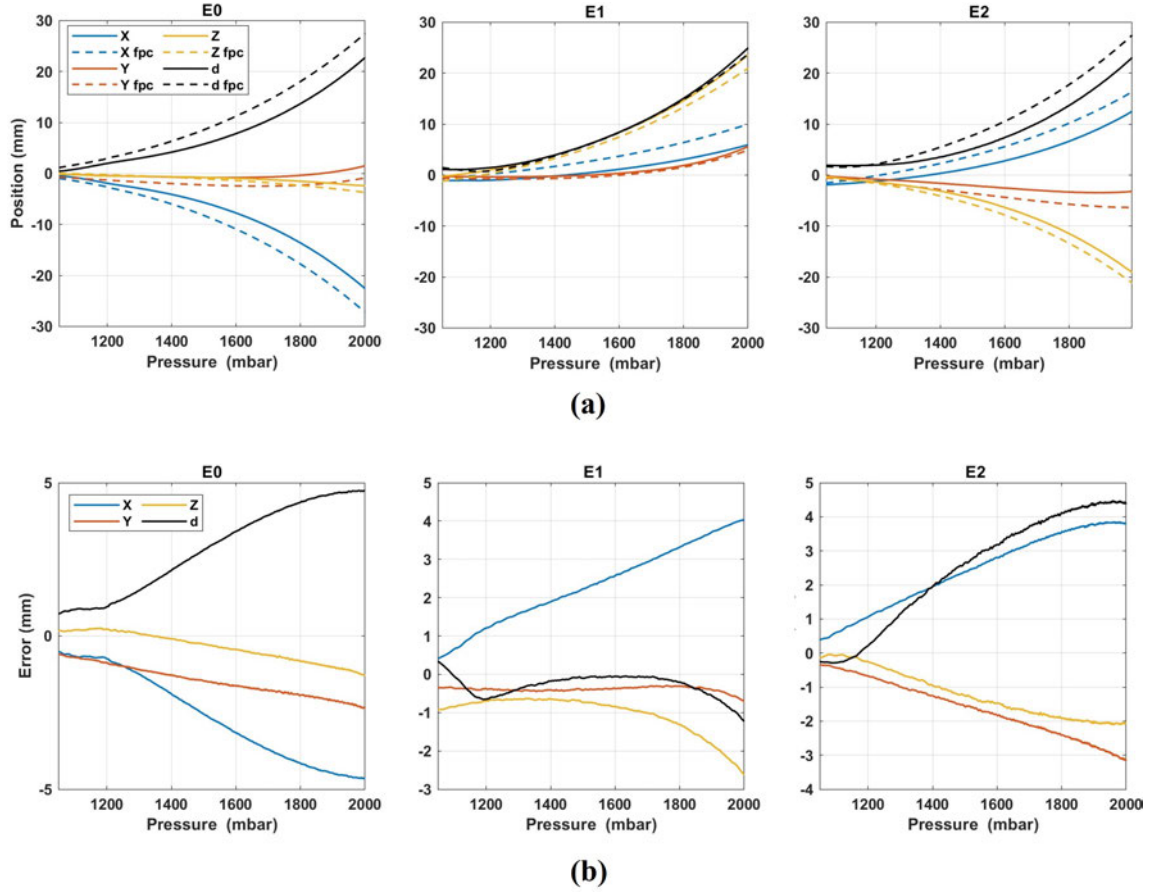


Figure 4.23: Tip trajectory alteration due to FPC integration in single chamber pressurisation: (a) Tip trajectory with and without FPC, (b) Divergence caused by FPC integration.

tions are more prominent than in y . The maximum displacements of the tip in x , z , and y are 49.61, 52.93, and 12.84 mm, respectively. The same phenomenon observed when analysing the stability of the EIT signals is also evident in Fig. 4.25(a), which demonstrates the single-chamber actuation of E1 using random sine-wave inputs. This actuation induces secondary pressurisation in the passive chambers. Specifically, at $t = 15.7$ s (vertical dashed line), a pressure change of 1100 mbar in E1 results in a 286 mbar increase in E0. At this timestamp, the inflation of E1 reduces its impedance by 1804Ω , while E0 exhibits an impedance increase by 827Ω due to compression (Fig. 4.25(b)).

Multi-chamber pressurisation shows combined chamber effects, with impedance–pressure trends unchanged from the single-chamber case. Between 28.6 and 33.5 s (vertical dashed lines in Fig. 4.25(c,d)), chamber E0 is deflating, leading to a reduction in E2 pressure, while E1 is inflating, exerting a compressive effect that increases E2 pressure. These opposing effects partially counterbalance each other, resulting in minimal net variation in E2 pressure. However, the E2 impedance changes are noticeable (Fig. 4.25(d)), with pressure remaining at 1143 ± 13 mbar while EIT values range within $4864.66 \pm 156.17 \Omega$. The

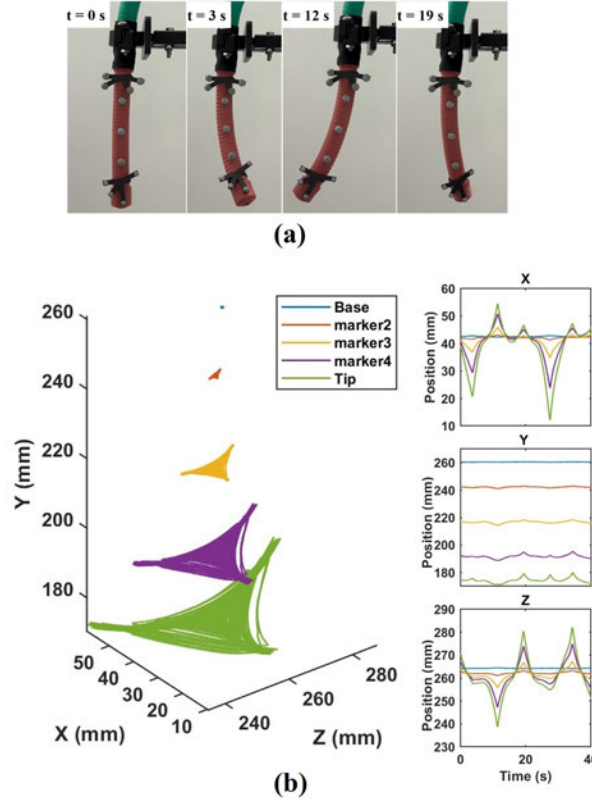


Figure 4.24: Free motion positional data acquisition with random pressurisation: (a) Ground truth via OptiTrack, (b) Markers' positional data in 3D.

maximum pressures applied to the SCM during free-motion data acquisition under both single- and multi-chamber pressurisation were 2.24, 2.31, and 2.24 bar for chambers E1, E0, and E2, respectively.

4.6.4 Reconstruction Results

Tip Trajectory

Fig. 4.26(a) and its 2D projections on different planes in Fig. 4.26(b) present the tip localisation results of the MLP network during free motion, evaluated on a test dataset containing 18,246 curves and motion instances. The dataset includes both single- and multi-chamber pressurisation-induced motions. Evaluations were performed on individual datasets as well as on the combined dataset to assess overall accuracy. All positional data used for prediction were baseline-calibrated with respect to the initial position of the base marker. The training and validation RMSE values were 0.05 mm and 0.07 mm, respectively, indicating low bias and variance during training with the combined dataset.

The prediction performance of the MLP for tip trajectory in Fig. 4.26 shows close track-

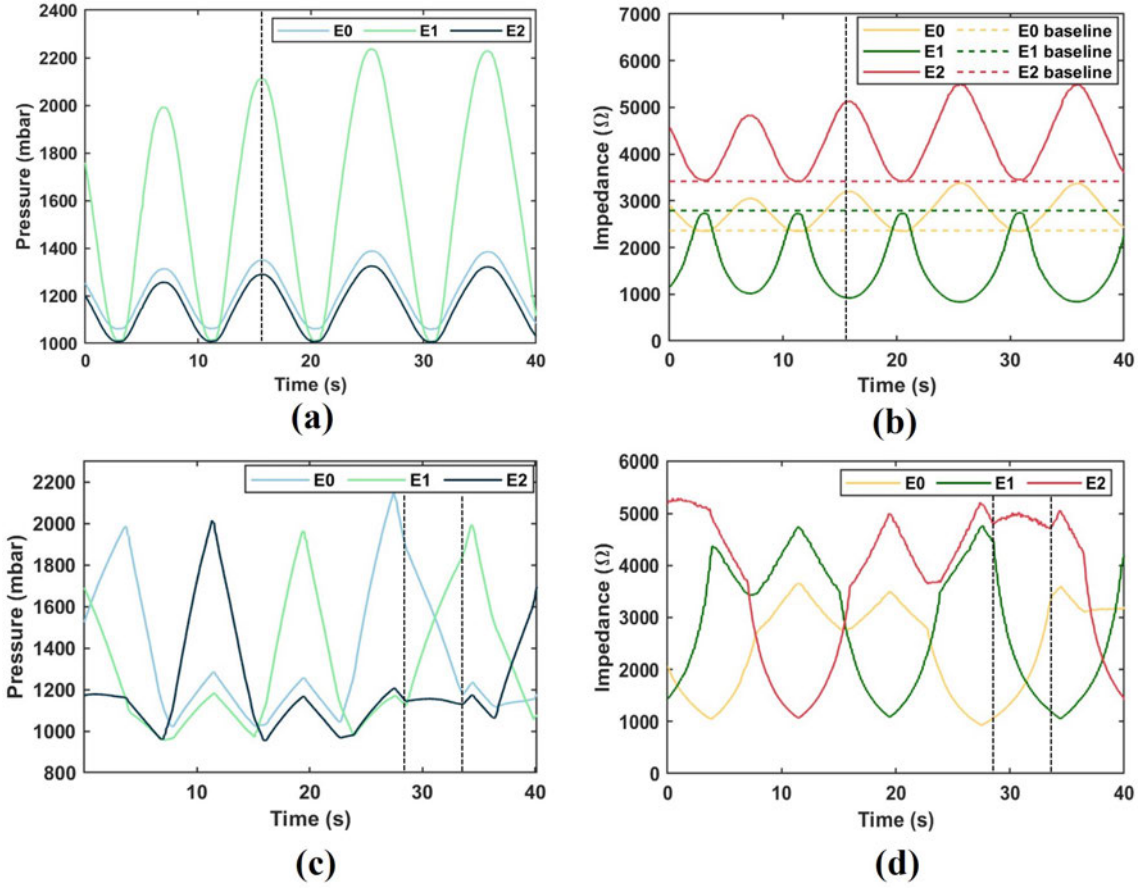


Figure 4.25: Free motion multi-modal data acquisition with random pressurisation: (a,b) Pressure and EIT (channel 5) raw signals in single chamber pressurisation, (c,d) Multi-chamber pressure and EIT (channel 5) raw signals.

ing of the generated tip path. The X–Z projections for selected trajectories, generated by single- and multi-chamber pressurisation, are shown in Fig. 4.27. The single-chamber case exhibits better estimation accuracy, attributed to reduced variability and the absence of compound effects from simultaneous chamber actuation compared to the multi-chamber case. In Fig. 4.27(a), the predicted trajectory diverges only in the upper branch of the path, starting at approximately $x = 10$ mm and $z = 20$ mm, while performing reasonably well elsewhere. In contrast, the multi-chamber trajectory in Fig. 4.27(b) shows discrepancies along almost the entire path, with the largest errors occurring at trajectory corners where the first chamber deflates and the subsequent chamber inflates. This suggests that the most significant prediction errors occur during transitions between active chambers. The higher error compared with the single-chamber case may also be attributed to the smaller number of motion samples in the dataset (65,537 versus 85,245), despite the greater variability present in multi-chamber actuation data.

The performance analysis by axis for single-chamber pressurisation of chamber E1, shown

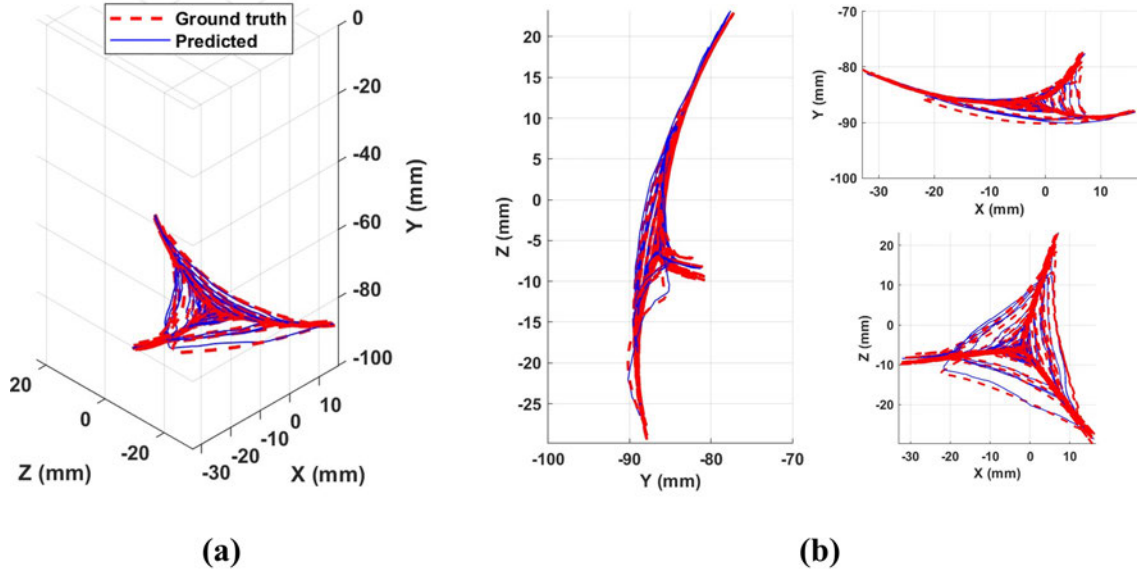


Figure 4.26: Tip localisation performance in free motion: (a) In 3D space, (b) 2D projections plots.

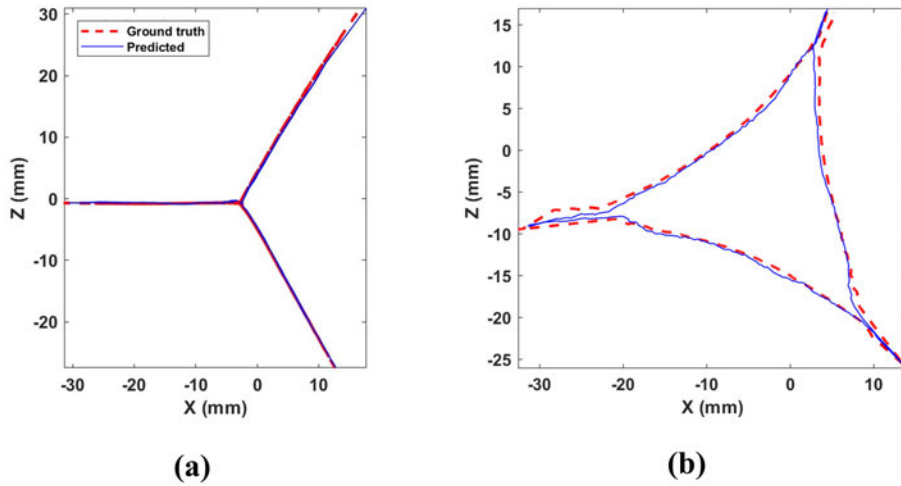


Figure 4.27: 2D projection of tip localisation on X-Z: (a) Single-chamber trajectory, (b) Multi-chamber Trajectory.

in Fig. 4.28, indicates that the predicted x , y , and z values are generally higher than the ground truth, resulting in estimation errors e_S calculated as Equation 4.14 and with a negative mean value.

$$e_S = S_{pr} - S_{gt} \quad (4.14)$$

where S is the 3D coordinate, and S_{pr} and S_{gt} are the predicted and ground truth values, respectively. The figure also shows that the x -axis error is the largest. The RMSE of tip estimation in single-chamber E1 pressurisation is 0.51, 0.29, and 0.35 mm for x , y , and z , respectively, based on predictions from 2,718 motion instances. RMSE and SD in single-

and multi-chamber pressurisations are highlighted in Table 4.5 and Table 4.6 with the corresponding number of motion instances noted.

The average RMSE across the coordinate frame for each chamber is computed as:

$$\text{Average RMSE} = \sqrt{\frac{\text{RMSE}_x^2 + \text{RMSE}_y^2 + \text{RMSE}_z^2}{3}} \quad (4.15)$$

Among all chambers, E2 yields the lowest average RMSE (0.21 mm), whereas E1 yields the highest (0.39 mm). For E0, this value is 0.35 mm.

Table 4.5: RMSE $\pm$ SD of positional error per axis for each active chamber in single-chamber pressurisation.

Active Chamber	$\mathbf{e}_x$ (mm)	$\mathbf{e}_y$ (mm)	$\mathbf{e}_z$ (mm)
E0 (2,720 curves)	0.53 $\pm$ 0.46	0.22 $\pm$ 0.21	0.18 $\pm$ 0.13
E1 (2,718 curves)	0.51 $\pm$ 0.38	0.29 $\pm$ 0.23	0.35 $\pm$ 0.32
E2 (3,021 curves)	0.21 $\pm$ 0.12	0.17 $\pm$ 0.16	0.24 $\pm$ 0.23

As discussed in previous sections on EIT characterisation, repeatability and robustness assessments indicated that E2 had the highest signal stability, while E1 had the lowest (see Fig. 4.17). In contrast, hysteresis was more favourable for E1. The average RMSE along each coordinate axis is 0.44, 0.21, and 0.26 mm for x , y , and z , respectively. These correspond to relative errors of 0.9%, 1.6%, and 0.5% compared to the maximum displacement in each direction within the training dataset (single- and multi-chamber combined).

Table 4.6: Tip estimation accuracy (RMSE $\pm$ SD) in multi-chamber pressurisation.

Error by axis (mm)	$\mathbf{e}_x$	$\mathbf{e}_y$	$\mathbf{e}_z$
Multi-chamber (9,787 curves)	0.48 $\pm$ 0.48	0.19 $\pm$ 0.18	0.49 $\pm$ 0.45

The overall tip estimation accuracy for the combined dataset (single- and multi-chamber) is summarised in Table 4.7, with an average RMSE of 0.37 mm. This corresponds to 0.35% of the total length of the SCM or 0.68% of the maximum tip displacement.

Table 4.7: Tip estimation accuracy (RMSE $\pm$ SD) over the combined dataset.

Error by axis (mm)	$\mathbf{e}_x$	$\mathbf{e}_y$	$\mathbf{e}_z$
Entire dataset (18,246 curves)	0.46 $\pm$ 0.30	0.20 $\pm$ 0.15	0.40 $\pm$ 0.26

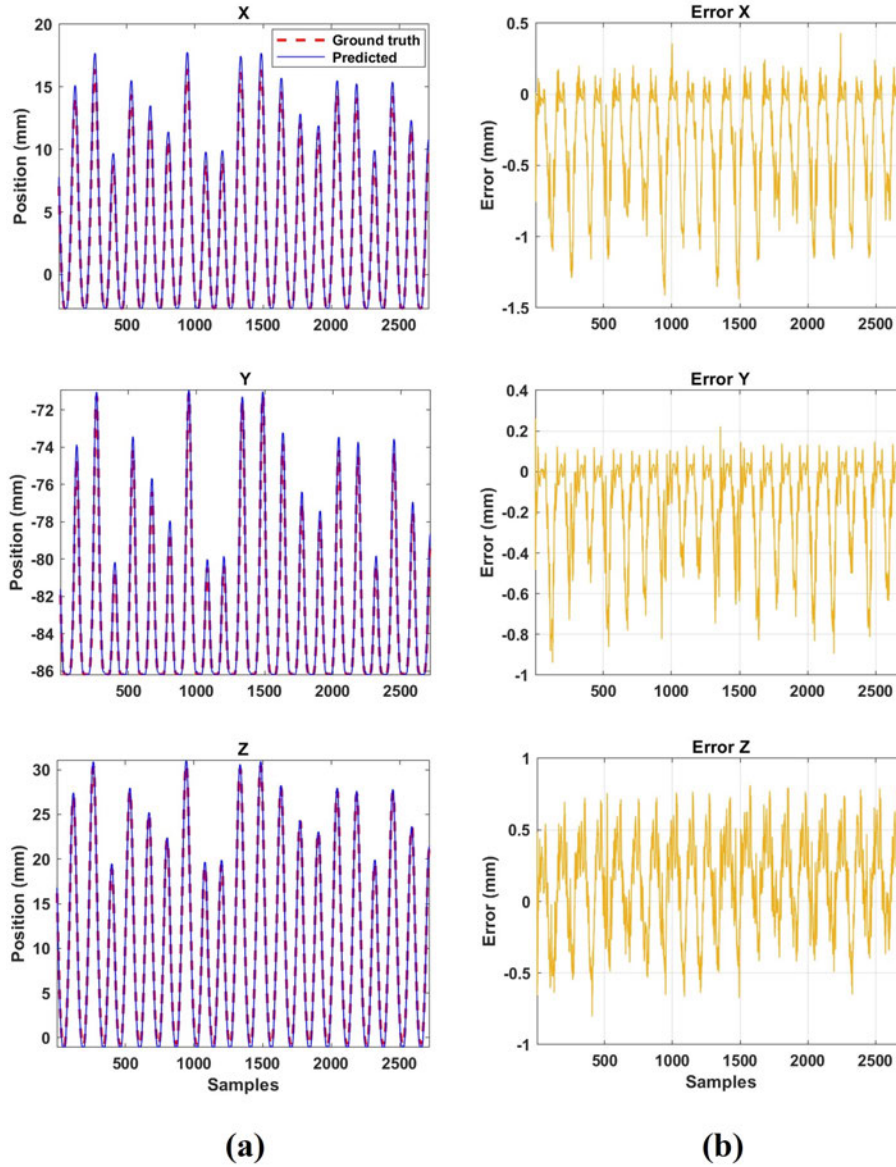


Figure 4.28: Tip localisation accuracy in single-chamber (E1) pressurisation: (a) Prediction vs ground truth, (b) Estimation error.

Backbone 3D Shape Reconstruction

The reconstruction accuracy was quantified by evaluating the tracking error of the keypoints and of 100 interpolated points distributed uniformly along the **SCM** backbone. For comparative analysis, the accuracy metrics were evaluated separately for the single- and multi-chamber pressurisation cases. The **RMSE** of the predicted keypoints is summarised in Tables 4.8 and 4.9.

The results indicate that, in general, the tracking error increases towards the tip of the **SCM**, consistent with the larger motion range observed in this region. An exception is

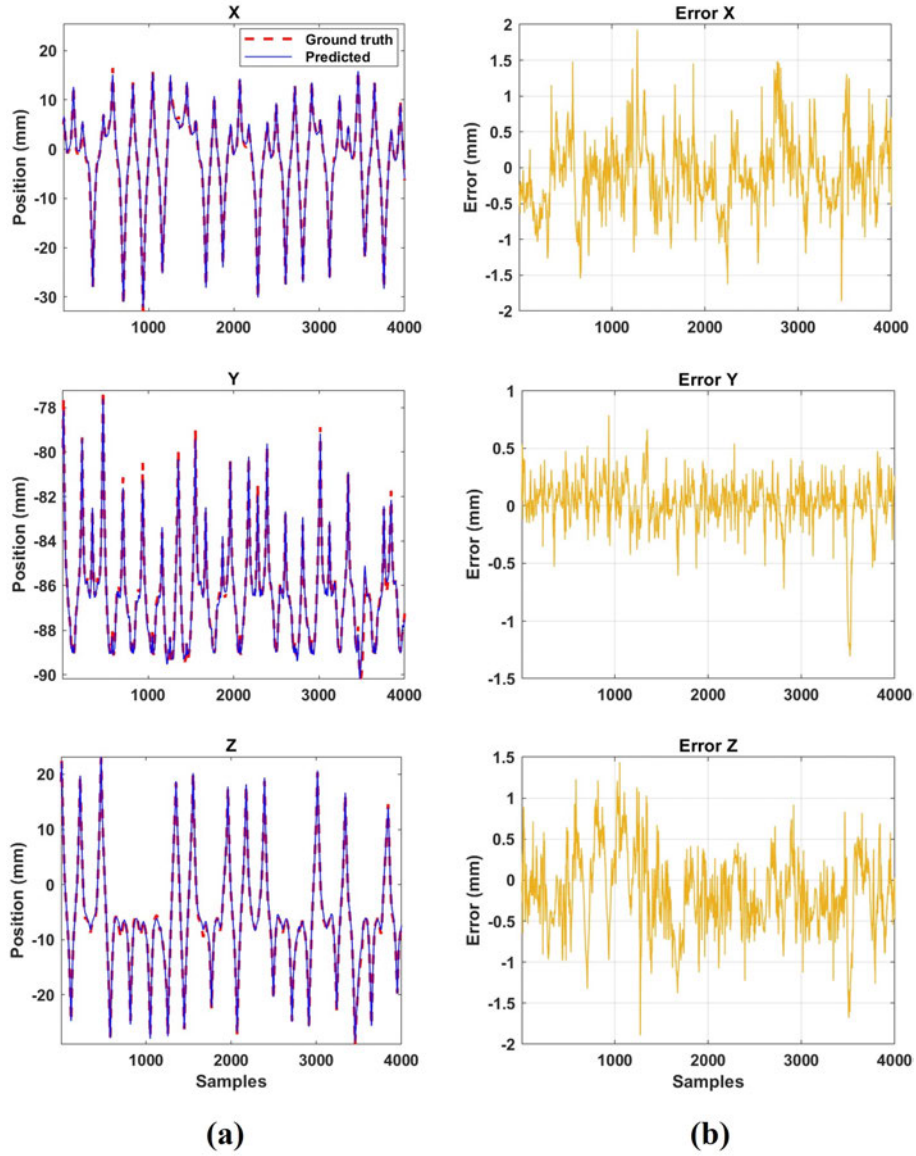


Figure 4.29: Tip localisation accuracy in multi-chamber pressurisation: (a) Prediction vs ground truth, (b) Estimation error.

Table 4.8: Key points estimation accuracy (RMSE $\pm$ SD) in single chamber pressurisation.

Marker location	e_x (mm)	e_y (mm)	e_z (mm)
Marker 2	0.06 ± 0.06	0.16 ± 0.15	0.13 ± 0.11
Marker 3	0.14 ± 0.11	0.09 ± 0.09	0.17 ± 0.16
Marker 4	0.26 ± 0.22	0.23 ± 0.23	0.22 ± 0.21
Tip	0.44 ± 0.35	0.21 ± 0.21	0.26 ± 0.26

noted in the single-chamber y-direction results, where Marker 3 exhibits a noticeably lower error than the others. This deviation may be attributed to the lower noise level in

Table 4.9: Key points estimation accuracy (RMSE $\pm$ SD) in multi chamber pressurisation.

Marker location	e_x (mm)	e_y (mm)	e_z (mm)
Marker 2	0.06 ± 0.05	0.10 ± 0.10	0.05 ± 0.05
Marker 3	0.16 ± 0.15	0.13 ± 0.12	0.14 ± 0.13
Marker 4	0.30 ± 0.29	0.18 ± 0.18	0.31 ± 0.29
Tip	0.48 ± 0.48	0.19 ± 0.18	0.49 ± 0.45

its positional data, as the marker is located near the midpoint of the [SCM](#), likely offering greater visibility to the cameras during data collection.

A comparison between single- and multi-chamber results reveals that, overall, the [RMSE](#) of keypoint predictions is lower for single-chamber pressurisation, with a few exceptions. In particular, in the y -direction, multi-chamber pressurisation yields lower errors for most markers due to the smaller working range in y .

The [RMSE](#) and [SD](#) of the pointwise accuracies, shown in Fig. 4.30, demonstrate a clear increasing trend in error from the base to the tip in both actuation modes. To better quantify the performance along the entire backbone, the [Mean Absolute Error \(MAE\)](#) (solid lines in Fig. 4.30) across the 100 interpolated points are presented in Table 4.10. These values show that single-chamber pressurisation produces lower [MAE](#) in the x - and z -directions, while multi-chamber pressurisation achieves lower errors in the y -direction.

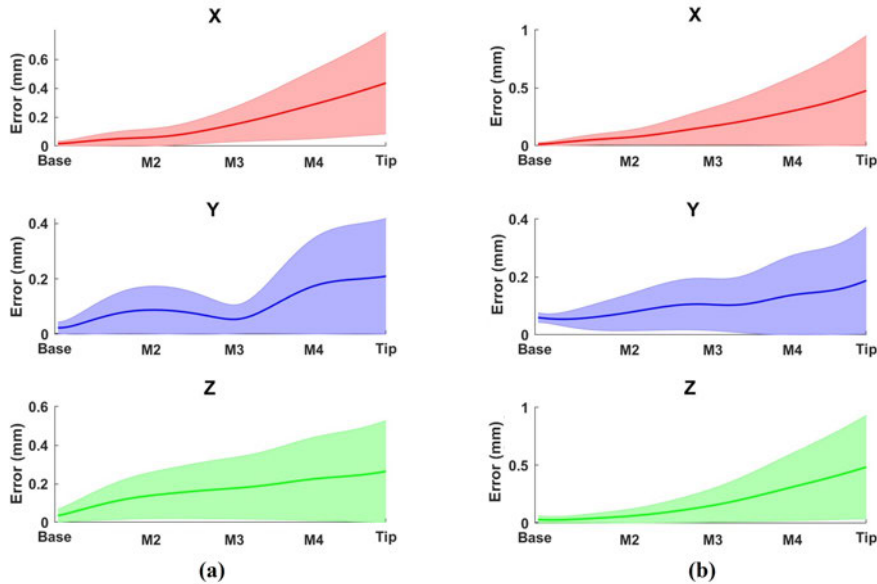


Figure 4.30: Progression of [RMSE](#) (solid line) and [SD](#) (filled area) of shape reconstruction from base to the tip: (a) Single-chamber, (b) Multi-chamber pressurisation. M2-M4 stand for markers two to four.

Table 4.10: Comparison of MAE between single- and multi-chamber pressurisation.

Actuation mode	e_x (mm)	e_y (mm)	e_z (mm)
Single chamber	0.1659	0.1037	0.1668
Multi chamber	0.1817	0.1034	0.1799

Finally, the backbone shape reconstruction results are visualised in Fig. 4.31, where cubic spline curves are fitted to both the predicted and ground-truth keypoints (including the interpolated points) for each actuation mode. Zoomed-in views of selected regions are also included to enable closer examination of local deviations between the predicted and actual curves, with more 3D curve samples illustrated in Fig. 4.32.

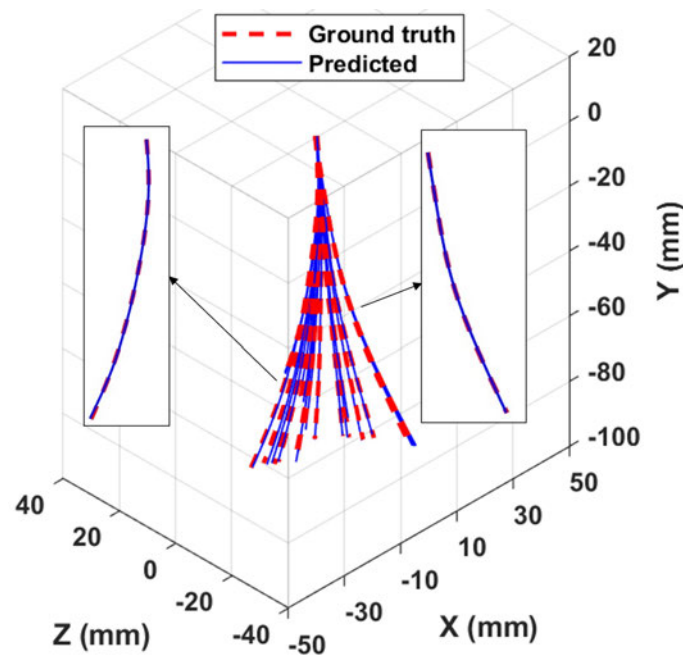


Figure 4.31: 3D shape reconstruction results in free motion for single- and multi-chamber pressurisation.

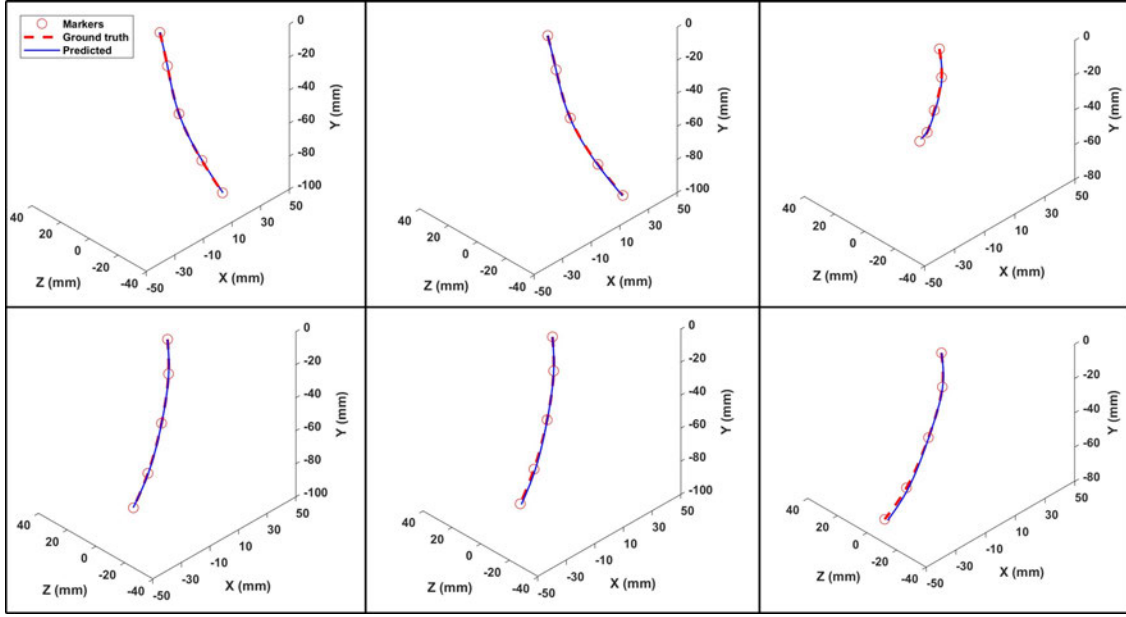


Figure 4.32: Sample 3D shape reconstruction results.

4.7 Discussion

In this chapter, the application of EIT proprioception in 3D for reconstructing the shape of a hydraulically actuated SCM using an MLP network was presented. The model was validated during free motion under both single- and multi-chamber pressurisation of the SCM. The resulting reconstruction errors demonstrate the promising applicability of EIT for proprioceptive sensing and acknowledge its potential as a feedback source for closed-loop control systems. The reported average localisation accuracy of 0.37 mm (RMSE, relative error 0.35%) demonstrates high precision, comparable to or surpassing existing sensing technologies. For instance, [125] reported a translation error of 0.9 mm in free motion by fusing resistive spring sensors with IMU data, effectively supporting closed-loop control. [233] achieved a mean tip localisation error of 0.83 mm (2.37% of the length) using FBG integrated into a 35 mm cable-driven surgical instrument. In another impedance-based study, [234] reported a minimum RMSE of 1.27 mm (0.42% of the length) using a combined LSTM and PCC model. However, their system employed a longer, three-segment soft robotic arm with a higher DOF than the SCM used in this chapter.

The length and other design parameters of the SCM were selected to maintain a diameter comparable to that of a conventional colonoscope, while the length was chosen to be consistent with previously developed soft actuators. Further optimisation of these parameters could enhance the flexibility of the SCM. Additionally, the chamber size and dimensions can be adapted to demonstrate the scalability of EIT and its applicability in other endo-

scopic procedures, such as bronchoscopy [44]. One limiting factor in the current design is the diameter of the endoscopic camera head; however, advancements in miniaturised camera technology will facilitate further reductions in **SCM** size (i.e., employing NanEye cameras (ams OSRAM, Austria) [235]).

During **SCM** fabrication, preventing leakage at the inlets is critical. In this work, the inlet tubes were made of materials other than silicone; therefore, a combination of silicone adhesive and mechanical knots was employed to ensure proper sealing. However, if the inlet tubes are replaced with silicone ones, applying a silicone-based adhesive (e.g., Sil-Poxy) alone would likely suffice due to its material compatibility with the **SCM**. Alternative approaches to prevent leakage can be explored as in [236].

Modifications in **FPC** design parameters, such as L_g and L_{f1} , can increase stretchability, although changes must adhere to manufacturer guidelines. There exists a mutual dependency between the **FPC** and **SCM** designs: altering the **FPC** width affects the **SCM** chamber width. For instance, increasing the number of electrode pads on the **FPC** necessitates design changes to accommodate additional wire traces. The minimum achievable trace width is constrained by manufacturing methods, which also influences the overall **FPC** width. Stretchability may be further enhanced by increasing the number of gaps or cuts in the kirigami pattern; however, manufacturing limitations such as laser cutter resolution impose restrictions.

Simulating **FPC** design using **FEA** can complement experimental evaluations, allowing detailed investigation of how design variations affect performance and the integrated **SCM** behaviour. Another important aspect to explore is the **FPC**'s life cycle and fatigue progression during repeated stretching and contraction cycles, where substrate cracking or degradation may cause sensor failure [237]. These investigations can be conducted both experimentally and through simulations.

Experiments assessing signal quality were performed with the **FPC** held in static configurations using 3D-printed fixtures. Extending these experiments to varying bending curvature or radius would enable comparison across diverse conditions. Such studies would elucidate the onset of plastic deformation in the **FPC** and the maximum strain limit before failure. Furthermore, the current simulations of the **FPC** within passive chambers considered only geometric effects of bending, neglecting the deformation induced by pressurisation of adjacent active chambers. This could be addressed by incorporating simulations where an active chamber adjacent to a passive one is pressurised, allowing more realistic deformation analyses.

The maximum and minimum hysteresis values of the **EIT** system at a maximum applied

pressure of 2.25 bar were measured as 20% and 11.7%, respectively. These values are comparable to those achieved by state-of-the-art shape sensing methods for colonoscopy; for example, capacitive sensors wrapped as sleeves around colonoscopes exhibit hysteresis ranging from 12.9% to 20.3% [238]. As demonstrated in this chapter, the intrinsic hysteresis of silicone rubber itself significantly contributes to the overall hysteresis of the EIT method. Further analyses using different silicone rubber materials could provide deeper insights into the role of hysteresis in EIT signals, thereby assisting in optimisation and customisation of the material to compose the SCM.

The polyamide substrate of the FPC inherently restricts the SCM's operational range at elevated pressures and large bending angles, despite efforts to increase stretchability. Replacing the polyamide substrate with an elastomer-based material, potentially integrated with graphene coatings or engravings [239], could enable multichannel EIT measurements while preserving full flexibility. Elastomer substrates, in addition to being ultra-flexible, offer enhanced durability by mitigating sensor failure. However, they may introduce additional sensing hysteresis or may impact the conductivity, hence necessitating meticulous investigations for optimised implementation.

The training procedure for the shape reconstruction estimator was restricted to axial calibration, in which the working channels were subjected only to tensile or compressive loading. While this assumption is adequate for initial proof-of-concept validation, it becomes invalid in more complex operational scenarios. During interaction of the SCM with surrounding tissues within the colon, additional external loads such as shear forces, torsional moments, and twisting wrenches are likely to arise. These loads may be applied both locally and continuously along the backbone, inducing complex chamber deformations that challenge the generalisation of datasets collected under free-motion conditions. Future studies should therefore acquire training data under experimentally realistic conditions, for example by manipulating the SCM within a phantom that replicates the confined colonic environment. Capturing EIT responses under such multi-axis, contact-driven loading conditions is expected to significantly improve the robustness and clinical generalisability of the proposed sensing approach.

Multiple experiments revealed discrepancies between baseline impedance values across the three chambers, as illustrated in Fig. 4.15, 4.20, and 4.25. Although identical baseline values are expected due to the symmetrical geometry and configuration of the chambers, fabrication-related variables likely contribute to these inconsistencies. In particular, the flexibility of 3D-printed mould materials such as Poly(lactic Acid) (PLA) may cause deformation in mould components due to the chambers' high length-to-width ratio, affecting parameters like wall thickness and initial chamber volume. Using stiffer materials in crit-

ical mould regions could yield more uniform chamber properties and reduce fabrication variability. Moreover, automating the manufacturing process, from silicone moulding to FPC affixation onto the working chambers, could improve fabrication repeatability and thus enhance consistency in EIT measurements.

The use of five optical markers distributed along the 106 mm SCM results in relatively large inter-marker distances, which inherently limits the spatial resolution of the reconstructed shape. Consequently, the fitted curves are derived from sparse discrete key points, reducing the fidelity of the reconstructed ground-truth geometry used to train the MLP. This limitation becomes more pronounced under complex deformation patterns, particularly when external forces induce localised curvature changes that cannot be adequately captured by sparsely distributed markers. Increasing the number of markers could improve spatial resolution and curve-fitting accuracy. However, excessive or inappropriately sized reflective markers may alter the morphology of the soft manipulator, introduce additional mass, or impose unintended gravitational loading, thereby affecting the mechanical behaviour being measured. Careful consideration is therefore required to balance measurement resolution against morphological integrity. Alternative approaches may provide more continuous shape information in future studies. For example, computer-vision-based methods can segment the manipulator backbone and track its deformation along the entire body, enabling continuous shape reconstruction rather than discrete point interpolation. Nevertheless, such techniques are susceptible to visual occlusion and limited field-of-view constraints, which are particularly relevant during *ex vivo* and *in vivo* trials.

Achieving submillimetre accuracy in tip localisation and shape reconstruction using the MLP underscores its suitability for inferring time-varying EIT readings. While more advanced network architectures could be explored, optimising hyperparameters such as the number of hidden units and layers, learning rate, and mini-batch size may yield improved performance with the current model. Ablation studies assessing the impact of individual EIT channels on prediction accuracy could further reduce computational cost, optimise electrode pad placement, and decrease the number of ports required on the Eliko system. This would free ports for additional FPC connections if the SCM were expanded beyond three chambers. To enhance the robustness and generalisation of the data-driven model, further strategies such as data augmentation (e.g., noise injection during training) and sim-to-real approaches [240] are recommended.

In the earlier chapters of this thesis (Chapters 2 and 3), LSTM architectures were employed for tip localisation and force estimation in planar SCM, owing to their suitability for modelling long-term dependencies in sequential data. However, the primary motivation for adopting a MLP in Chapter 4 for three-dimensional reconstruction is to establish

the foundation for real-time implementation of the EIT sensing framework, particularly within a closed-loop control architecture. Compared with LSTM networks, MLPs exhibit lower computational complexity, enabling faster training and inference, improved convergence behaviour, and greater flexibility with respect to training dataset size. Furthermore, their reduced architectural complexity simplifies hyperparameter tuning relative to recurrent models [241]. Although LSTMs are capable of capturing temporal dependencies, achieving high accuracy and close agreement with ground-truth data in certain applications often necessitates architectural modifications, which further increase model complexity [242]. Notably, a comparative study on predicting propulsive forces in an underwater soft robot reported that MLPs achieved higher accuracy and more efficient training than LSTMs, particularly as network depth increased [243]. In light of these considerations, the MLP architecture was adopted in this chapter, replacing the LSTM-based models utilised in Chapters 2 and 3.

Given the size of the training dataset and the dimensionality of the input and output spaces (18 and 15, respectively), the MLP with the selected architecture required 184 s for training on a single GPU. In contrast, the LSTM network employed for planar tip localisation in Chapter 2 required 532 s under the same hardware configuration. Notably, in the planar localisation experiment, the input and output dimensions were 9 and 3, respectively. Furthermore, the planar dataset comprised 14,042 instances, which is less than one-eighth of the dataset used for three-dimensional shape reconstruction (114,142 motion samples). Despite the lower dimensionality and smaller dataset size in the two-dimensional case, the LSTM required approximately 2.89 times longer training time than the MLP. This computational advantage of the MLP is particularly relevant for online learning paradigms, where models must be updated or fine-tuned continuously from streaming data rather than fixed batches [244]. These findings further support the suitability of MLP-based architectures for real-time deployment of data-driven shape reconstruction compared with RNN-based approaches.

Despite the limitations identified, assessing whether the current system can achieve compelling performance in clinical settings through *ex vivo* and *in vivo* trials remains a critical avenue for future work. This pursuit, followed in the next chapter, is motivated by the promising results presented herein, including accurate shape reconstruction and the demonstrated repeatability and robustness of the EIT signal.

Chapter 5

EIT Evaluation in Clinical Practice

5.1 Introduction

Developing shape sensors that are compatible with soft endoscopes and robust under clinical conditions remains a significant challenge. Most existing shape-sensing modalities either compromise system integrity or lack comprehensive validation in clinical settings. Consequently, translating sensing technologies from controlled environments to real-world clinical scenarios is crucial for advancing their practical applicability. A prerequisite for such translation is the identification of the current technical challenges and assessment of how the proposed approach can mitigate them.

Current challenges in translating sensing technologies from bench-top experiments to clinical examination typically arise from incompatibility with the existing conditions in the operating theatre, as well as the complexities associated with operating on live models or on tissue phantoms. [FBGs](#), for instance, are brittle and may not capture complex shapes accurately due to both their limited flexibility and constraints on the number of sensors that can be embedded into the device [\[245\]](#). These limitations can disrupt their functionality during procedures and lead to failure. [EM](#) trackers, in addition to being susceptible to magnetic field distortions, have limited spatial coverage. Image-guided sensing and control can also introduce complications such as inefficient data acquisition pipelines, elevated radiation dose, high equipment cost, and workflow interruptions [\[246\]](#).

Miniaturisation is another key requirement for clinical practicality for soft robots [\[247\]](#), particularly when integrating additional sensing modalities. For example, [FBGs](#) often require a separate channel, as do other sensing methods. Even if miniaturisation is feasible, the overall size constraints of the device may necessitate downscaling of other essential

components, such as the camera, biopsy tools, or irrigation systems, which can increase manufacturing costs due to the precision required at smaller scales.

Dynamic environmental effects and their uncertainty often restrict the clinical validation. Interaction between the instrument and tissue induces external forces stemming from friction. Temperature variations inside tissue or a live model can also influence sensor readings. Respiratory-induced motion may introduce additional challenges, particularly if the measurement system is overly sensitive to such movements, leading to errors in the prediction phase. Furthermore, intra-anatomical pressures, such as those within the colon or other endoluminal tracts, can impact readings, especially in systems employing soft robotic mechanisms [248]. These factors necessitate rigorous evaluation under realistic surgical conditions beyond bench-top validation to fully characterise environmental impacts.

Several studies have investigated tip localisation and shape reconstruction for surgical instruments in endoscopy under clinical and preclinical adjustments, each employing different validation methods and applications. ScopeGuid (Olympus, Tokyo, Japan) integrated with EM sensors was used in [249] to perform colonoscopy on patients and to evaluate their overall satisfaction in terms of sedation use. In [250], an EMT system was integrated within the working channel of an endoscope and tested during endoscopic necrosectomy both with and without the endoscopic camera. The *ex vivo* setup comprised a 3D-printed phantom of a stomach and pancreatic necroses. In [251], optimal C-arm orientation for adaptive intraoperative imaging-based localisation was studied. The method was validated using a silicone aortic arch phantom and a robotic catheter, with fluoroscopic and CT imaging as ground truth. Reconstruction involved segmentation of the catheter's central line followed by 3D estimation, view orientation optimisation, and localisation, with accuracy assessed via mean point-to-centreline distances relative to CT-derived ground truth. In [252], a pneumatically actuated soft sleeve incorporating five optical sensors was developed to enhance safety during colonoscopy while enabling shape sensing. Fabricated from Ecoflex 00-30 and Dragon Skin 30, the sleeve was compatible with a 12 mm OD conventional endoscope, reaching 23 mm after mounting. A decoupling algorithm differentiated between orientation, curvature, and applied force. Trials were conducted on a 60 cm bovine colon for *ex vivo* testing and on two 30 kg porcines for *in vivo* validation, with additional analysis on its effect on learning curves and caecum intubation time. Given the mechanical, electromagnetic, and cost limitations of current clinical sensing technologies, a functional and affordable alternative is required.

Building on the accuracy achieved in free-motion environments in the previous chapter, the present work evaluates the performance and robustness of the proposed sensing

method under clinical conditions via *ex vivo* and *in vivo* trials. Specifically, the ability of EIT combined with an MLP network to reconstruct the shape of the SCM is assessed in complex and unpredictable surgical environments. The same ANN architecture is used, trained on multimodal data comprising pressure, EIT, and positional measurements. A porcine colon tissue model is employed for data collection in the *ex vivo* phase. Subsequently, an *in vivo* study is conducted using a 35 kg porcine model to evaluate usability, reliability, and accuracy.

Under *in vivo* conditions, a range of uncertain factors are expected, including contact friction between the instrument and tissue, instrument twisting, tissue motion, and compression forces induced by the colonic wall movement. Evaluating these effects and compensating for them using the current SCM design and data-driven model remains challenging. Although these factors are not explicitly incorporated into the estimation model, the objective of the *in vivo* studies is to establish an understanding of SCM deployment and to assess the feasibility of capturing deformation-induced EIT data, while observing the complications involved in validating such an approach. It is worth noting that, due to existing challenges associated with inserting the SCM, validating the data and the inferences presented in this section against alternative approaches in the literature requires substantial further investigation.

The key objective of *ex vivo* and *in vivo* studies is to inspect the performance of EIT reconstruction under the aforementioned technical challenges. Certain challenges, such as miniaturisation, can potentially be addressed by the EIT approach, as the overlap between the actuation and sensing media offers the possibility for more compact integration, thereby facilitating clinical adaptation. However, dynamic uncertainties and other environmental factors must be examined thoroughly.

The author contributed to the experimental design, hardware preparation, setup assembly, coordination of personnel and clinical equipment, data collection, and subsequent analysis and prediction. These clinical trials were conducted during a five-month research exchange at the MRC in Hong Kong, a collaborator and sponsor of this study.

The Methods section details the experimental hardware and configurations for both *ex vivo* and *in vivo* trials. Furthermore, the compatibility of EIT with clinical environments, including signal integrity under X-ray and Magnetic Resonance Imaging (MRI) exposure, is discussed. The acquired datasets are then analysed, and prediction results from the MLP are presented alongside shape reconstruction outcomes for the *in vivo* trials in the Results section. Finally, the Discussion section addresses potential avenues for improvement and examines the limitations of applying this method in clinical practice.

5.2 Methods

The control system remains essentially the same as described in the previous chapter, with minor modifications for registering positional data. The EIT protocol follows the procedure outlined in Table 4.3, with measurements recorded at a frequency of 61 kHz. The hardware setup in Fig. 4.8(a) provides actuation, sampling, and transmission of both EIT and pressure data. The same MLP architecture shown in Fig. 4.7 is trained on the data collected in *ex vivo* trials.

5.2.1 Clinical Compatibility Tests

Prior to the *ex vivo* data collection, a series of examinations was conducted to evaluate the output visualisation of EIT components under a 1.5 T MRI machine (SIGNA Artist, GE HealthCare, USA), as well as to assess potential interference from X-ray imaging. The objective was to confirm that EIT operation remains valid in clinical settings.

For MRI test, two SCMs, one with the FPC integrated and filled with saline and one without, were placed on the patient table. The control system and the overtube were positioned away from the machine. A water-filled bottle was placed nearby as a reference. After considering the safety concerns, the arrangement was then introduced into the bore, as shown in Fig. 5.1, and MRI images were captured to highlight the difference between the visualisation of an SCM with and without FPC.



Figure 5.1: Evaluating MRI imaging of EIT components (excluding Quadra).

For X-ray compatibility, EIT signals were recorded with the SCM positioned under the

machine, and the consistency of the measurements was assessed to determine whether X-ray exposure introduced distortions.

5.2.2 *Ex vivo*

Experimental Setup

Ex vivo experiments are conducted using porcine colon tissue (see Fig. 5.2(c.II)), measuring 650 mm in length with a diameter reducing from 40 mm at the rectum towards the caecum. A trocar and a height-adjustable frame at the entry point (Fig. 5.2(b)) allow endoscopic insertion. Unlike free-motion trials, where optical motion capture was feasible, insertion inside the colon tissue or later into an animal model precludes the use of such systems. Instead, positional data are acquired using the Aurora EMT system (NDI, Canada), as illustrated in Fig. 5.2(c.I).

Three EM sensors (0.5×8 mm, five DOFs each) are mounted on the SCM via silicone belts fabricated from elastosil, the same material as the SCM, ensuring flexibility and structural integrity. Two sensors are positioned at the base and tip, at similar locations to the OptiTrack markers used previously, while the third is placed mid-length, corresponding to Marker 3 from the free motion setup. The limited number of sensors is due both to hardware constraints (availability of sensors and SIU (Sensing Interface Unit) slots) and the aim of reducing wiring complexity.

The colon tissue is housed within a plastic enclosure arranged to replicate the rectum–sigmoid pathway. The enclosure is positioned above the EM field generator and secured with 3D-printed supports. An EVIS X1 colonoscope (Olympus, Japan) (Fig. 5.2(a)) is available for insufflation when required. An integrated endoscopic camera within the SCM provides intraluminal visual feedback, verifying tip deflection during insertion. Although not used for training or shape reconstruction, additional visualisation is provided by a robot-assisted C-arm system (Artis Zeego, Siemens), also shown in Fig. 5.2(a), which offers intraoperative 2D X-ray imaging for illustrative purposes.

To maintain consistency with the free motion experiments in Chapter 4, positional data obtained from both *ex vivo* and *in vivo* studies are transformed into the same coordinate frame. Beyond this unification, monitoring the rolling angle of the endoscope during insertion is essential for navigation. Occlusion by tissue or the animal model prevents direct visualisation of orientation, making it difficult to determine which chamber’s activation directs the tip towards the lumen centre. Visual feedback during clinical trials from the integrated endoscopic camera and X-ray imaging provides 2D projection views, which lack roll-angle information.

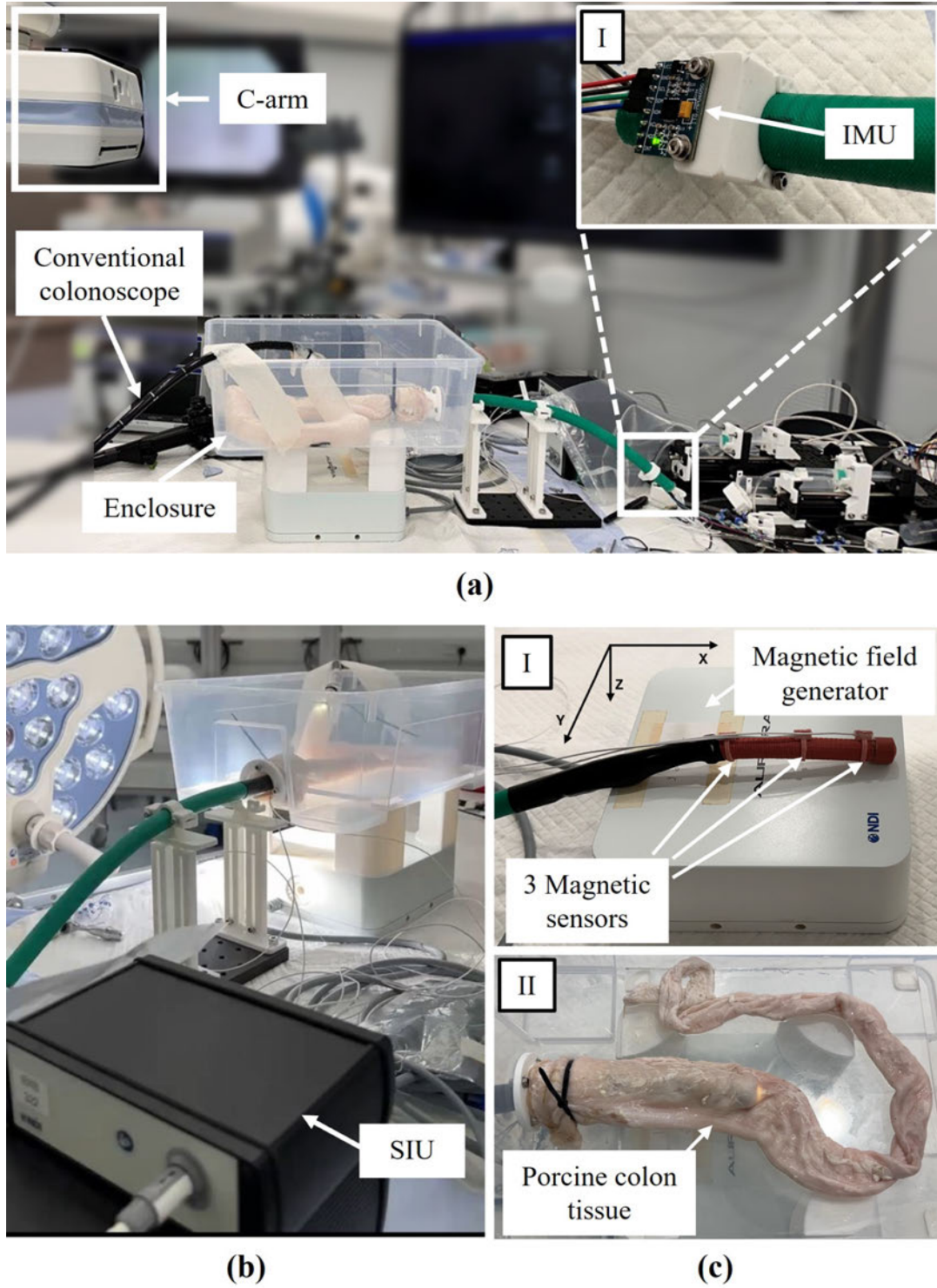


Figure 5.2: *Ex vivo* experimental setup: (a) Overview of the actuation and acquisition units with zoomed-in view of IMU (I), (b) SCM held by a height-adjustable frame and inserted through a trocar, (c) Aurora and EM sensors (I) with porcine colon arranged in 2D (II).

To address this, an IMU is mounted at the proximal end of the flexible overtube, where hydraulic tubes and wiring exit to connect with the actuation and sensing unit (see Fig. 5.2(a.I)). The overtube is resistant to torsion along its central axis, enabling the IMU to accurately reflect the SCM's roll angle for navigation purposes.

Data Acquisition

As the colon tissue was configured in a plane, only one chamber was activated at a time, with data collected under random pressure amplitudes. Before switching the active chamber, the overtube was rotated to align the active chamber's bending plane with that of the tissue, thereby minimising misalignment, while the rotations were simultaneously recorded by the IMU. It should be noted that due to interference between X-ray radiation and the EMT system, dataset collection was performed solely with the Aurora system operating, while X-ray imaging was deactivated. X-ray was later employed to capture the continuous shape changes of the SCM.

In total, 74,717 motion samples were collected at an average sampling rate of 6.3 Hz using the Aurora system. Of these, 20,899 curves correspond to E0 activation, 30,281 to E1 activation, and 23,537 to E2 activation, with data collection times of 26.2, 37.8, and 29.4 minutes, respectively. The dataset was subsequently divided into training, validation, and test subsets in a 70:15:15 ratio, consistent with the procedure applied to the free motion dataset.

5.2.3 *In vivo*

Experimental Setup

The control system used for *ex vivo* data collection was also employed for *in vivo* trials in Fig. 5.3(a.I). However, instead of Aurora and EM sensors for positional ground-truth data, only X-ray imaging was used to capture shape changes and subsequently validate reconstruction predictions. The wiring of the integrated endoscopic camera within the SCM produced sufficient contrast in the X-ray images, enabling the backbone shape of the SCM to be visualised. The EMT system was omitted because, during *in vivo* experiments, the SCM was subjected to unpredictable and substantial external forces from colonic contractions, which could displace or detach EM sensors mounted on the device. Furthermore, positioning the animal above the EM field generator risked exceeding its operational range.

The animal model was a 35 kg porcine, anaesthetised in the preparation room by an expert anaesthesiologist before being transferred to the operating room and positioned on

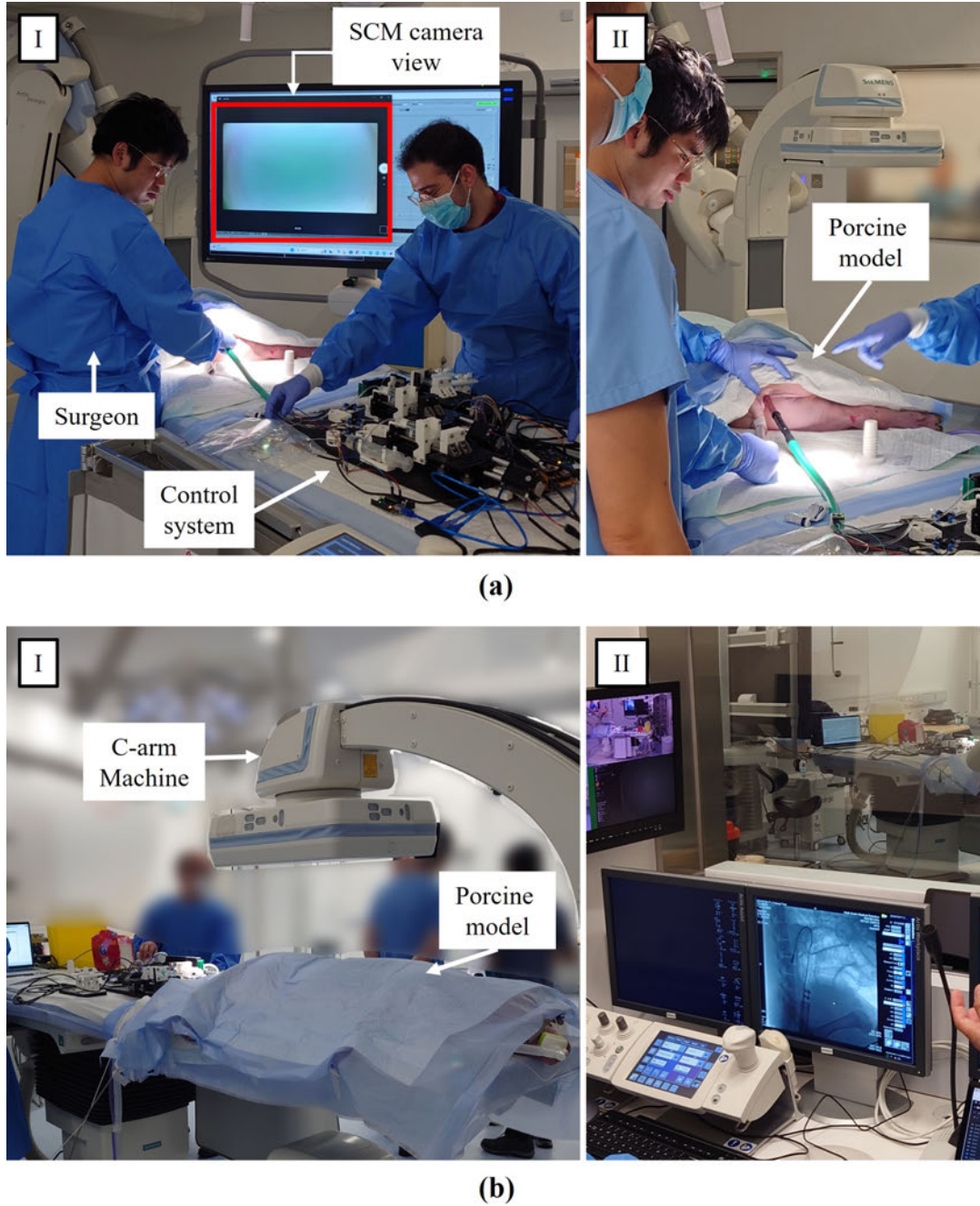


Figure 5.3: *In vivo* trial: (a) Control system and the surgical team (I) introducing the SCM through the porcine's anus (II), (b) Animal model positioned under X-ray imaging (I) and capturing the ground truth shape from the control room (II).

the surgical table within the X-ray field, shown in Fig. 5.3(b.I). Vital signs, including heart rate and oxygen saturation, were continuously monitored throughout the procedure. The trial was supervised by Dr. Hon Chi Yip, an expert gastroenterologist. All ethical and experimental protocols were approved by the Institutional Animal Care and Use Committee (IACUC) of the Hong Kong Science and Technology Parks (HKSTP).

Data Acquisition

Data collection began with the insertion of the **SCM** and the EVIS X1 into the porcine via the anus (as in Fig. 5.3(a.II)), while the gastroenterologist navigated using visual feedback from both the integrated **SCM** camera and the conventional colonoscope. The initial objective was to advance the endoscope approximately 400 mm into the colon without recording data from any multimodal sensing systems. Data acquisition commenced during retraction.

As X-ray imaging was the sole modality for capturing the **SCM** shape, and given safety concerns associated with radiation exposure, simultaneous retraction and continuous imaging were not feasible. During X-ray operation, the surgical team relocated to the control room in Fig. 5.3(b.II). Consequently, **EIT**, pressure data, and X-ray images were acquired at discrete static configurations, where the **SCM** remained stationary and chamber pressures were held constant.

The protocol for each configuration followed five steps: (1) after intubation, 50–70 mm of the **SCM** was retracted, (2) **EIT** and pressure data recording commenced, (3) clinical staff relocated to the control room, (4) X-ray footage was captured, and (5) **EIT** and pressure recording was terminated. The datasets obtained at these stages were subsequently employed as the test set for the **MLP** model trained on the *ex vivo* dataset. Particularly, the *in vivo* data remained entirely unseen during training.

5.2.4 Data Post-Processing

The *ex vivo* dataset was baseline-corrected after synchronisation and interpolation of the sensing modalities using **PCHIP**. Positional data were represented with respect to the base **EM** sensor. As described earlier regarding coordinate frame unification, the positional data obtained in clinical trials were transformed into the free motion coordinate frame, as shown in Fig. 4.8(b). This transformation was applied through rotational matrices, as expressed in Equation 5.1.

$$S_f = \left(\prod_{i=1}^3 Q_i \right) \cdot S_a^T \quad (5.1)$$

Here, Q denotes the rotational matrices, with $Q_1 = Q_z(\alpha)$, $Q_2 = Q_y(\beta)$, and $Q_3 = Q_x(\phi)$. The angle ϕ represents the roll angle recorded by the **IMU**, while $\alpha = \pi/2$ and $\beta = \pi$. S_a represents the positional data in the Aurora coordinate system arranged according to Equation 4.4, whereas S_f denotes the corresponding transformed data in the free motion

coordinate system. The resultant S_f values were then used in Eqs. 4.3 and 4.4.

5.3 Results

5.3.1 *Ex vivo*

Data Acquisition

The view from the endoscopic camera integrated into the **SCM** and the corresponding X-ray images from the C-arm in Fig. 5.4 illustrate **SCM** deformation. The X-rays also show the EVIS X1 used for tissue insufflation. The endoscopic view confirms **SCM** bending and tip motion through the shift in the centre of the colon tract from bottom-right in frame (1) towards the centre in frame (4). **SCM** deflection is further visible through the camera wiring in the X-rays, with the maximum deflection observed in frame (4). The complete single inflation–deflation cycle lasted 10.2 s.

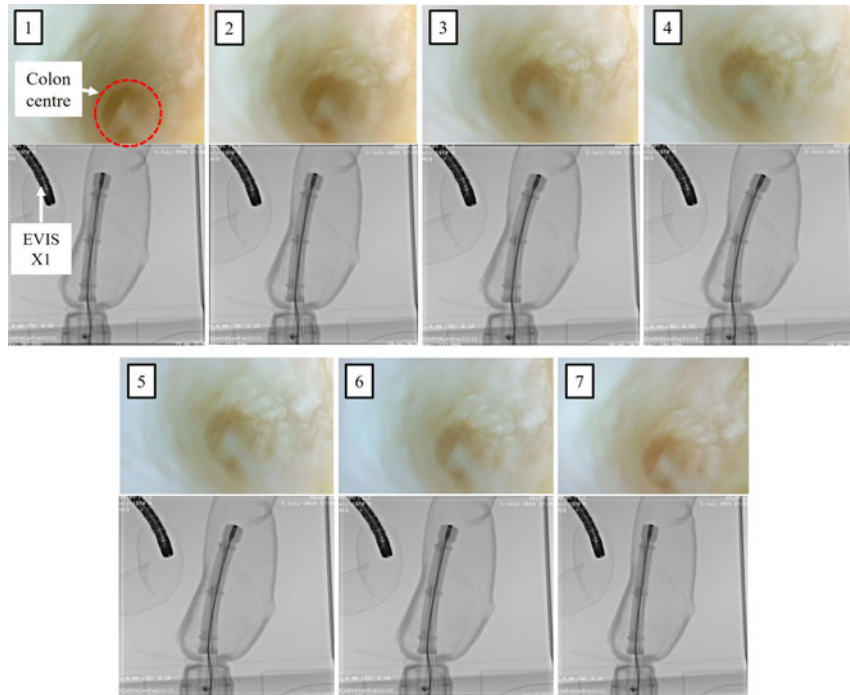


Figure 5.4: **SCM** camera view and corresponding X-ray images over seven frames, illustrating one full cycle of inflation (top) and deflation (bottom) in *ex vivo*. EVIS X1 is also captured in the top left corner of X-ray frames.

Positional data from the three **EM** sensors in **3D** (Fig. 5.5(a)), after coordinate transformation according to Equation 5.1, indicate a more discrete workspace compared with free motion. For the transformation, **IMU** readings yielded roll angles of 0° , 120° , and 250°

for activations E0, E2, and E1, respectively, with E0 serving as the reference and the other values representing counter-clockwise rotations.

The maximum Euclidean displacement of the SCM tip was 28.63 mm during E1 activation, equivalent to 52% of the free-motion range. For other chambers, maximum tip displacements were 23.46 mm (E0) and 28.41 mm (E2). The maximum displacement of the middle sensor was 8.65 mm during E1 activation (30% of the tip displacement), with E0 and E2 yielding 7.92 mm and 7.93 mm, respectively (28% of the tip displacement). After transformation into the free-motion coordinate frame, maximum axis-specific tip displacements were 21.01 mm along the x-axis (E0), 15.27 mm along the y-axis (E2), and 26.3 mm along the z-axis (E1) (see Table 5.1).

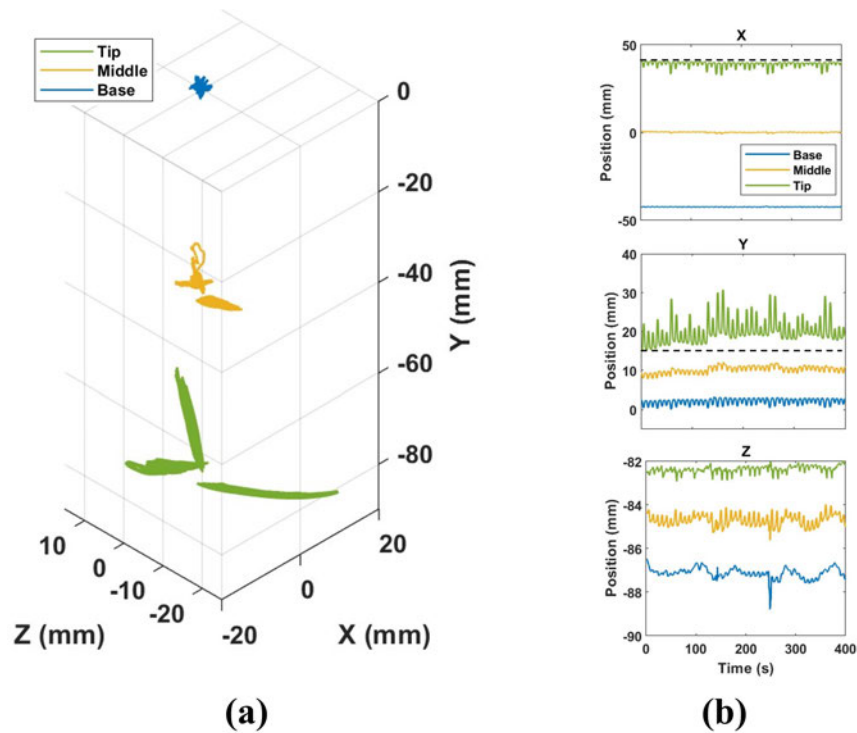


Figure 5.5: EM sensors trajectory in *ex vivo*: (a) 3D representation of transformed positional data, (b) Trajectories per axis before coordinate transformation.

Table 5.1: Maximum tip displacements per axis in the transformed coordinate frame.

Axis	Maximum displacement (mm)	Activation
x	21.01	E0
y	15.27	E2
z	26.30	E1

Axis displacements before transformation and baseline correction (Fig. 5.5(b)), relative to the coordinate frame defined in Fig. 5.2(c.I), show that upon pressurisation, x-values de-

crease from baseline (dashed lines), whereas y-values increase. Differences in x-positions across sensors reflect their relative separations along the **SCM**. Baseline offsets in the y-axis arise from the pre-bent **SCM** configuration, as also evident in the X-ray image in frame (1) of Fig. 5.4. Instead of being straight, the **SCM** exhibits a slight slope. Along the z-axis, deflections remain nearly constant, with variations across sensors attributable to **SCM** inclination caused by gravity and the misalignment between the trocar entry point and the tissue plane.

Over time, the y-tip trajectory (green line in Fig. 5.5(b)) shows an offset between its baseline (horizontal dashed line) and deflated rest positions, indicating incomplete return due to tissue friction. A similar offset is observed in the x-axis, though less pronounced. The largest offsets were 12.28 mm (y-axis) and 4.04 mm (x-axis), both during E0 activation. However, this offset is not reflected in the **EIT** data (Fig. 5.6(b)), as the active chamber is equally expanded as in the initial rest state, yielding comparable impedance measurements.

Fig. 5.6 shows the pressure and raw impedance data from channel 5 of each chamber. A full inflation–deflation cycle (highlighted from 100 s onward) is shown on the right side of each plot. During E0 activation, passive chamber pressures increased proportionally but remained smaller than the active pressure. Impedance in the active channel decreased, while passive impedances increased, consistent with free-motion results. In the zoomed view, a 1265 mbar increase in E0 pressure corresponded to a decrease of 818.87 Ω in channel 5 impedance. For passive chamber E1, the respective changes were a 298 mbar pressure increase (24% of the E0 pressure change) and a 705.21 Ω rise in impedance (86% of the E0 impedance change). Maximum pressures reached 2.67, 2.56, and 2.63 bar for E0, E1, and E2, respectively. Despite higher pressures compared with free motion, *ex vivo* displacement was restricted, highlighting the frictional constraint imposed by tissue contact.

Clinical Compatibility Tests

During the X-ray compatibility test, chamber E1 was pressurised with random amplitudes, as illustrated in Fig. 5.7(b), while both **EIT** and **EM** data were recorded. At approximately 562 s from the start of data collection, the X-ray machine was switched on. This moment, indicated by the dashed vertical line in Fig. 5.7(a), corresponds to a sudden distortion in the y-coordinate of all three **EM** sensors. Prior to this point, the y-coordinate of the tip sensor, for example, fluctuated in response to the applied pressure. Once the X-ray was activated, however, the **EM** sensor values exhibited an abrupt jump or drop and then remained constant despite continued dynamic pressurisation of the chamber. This distortion

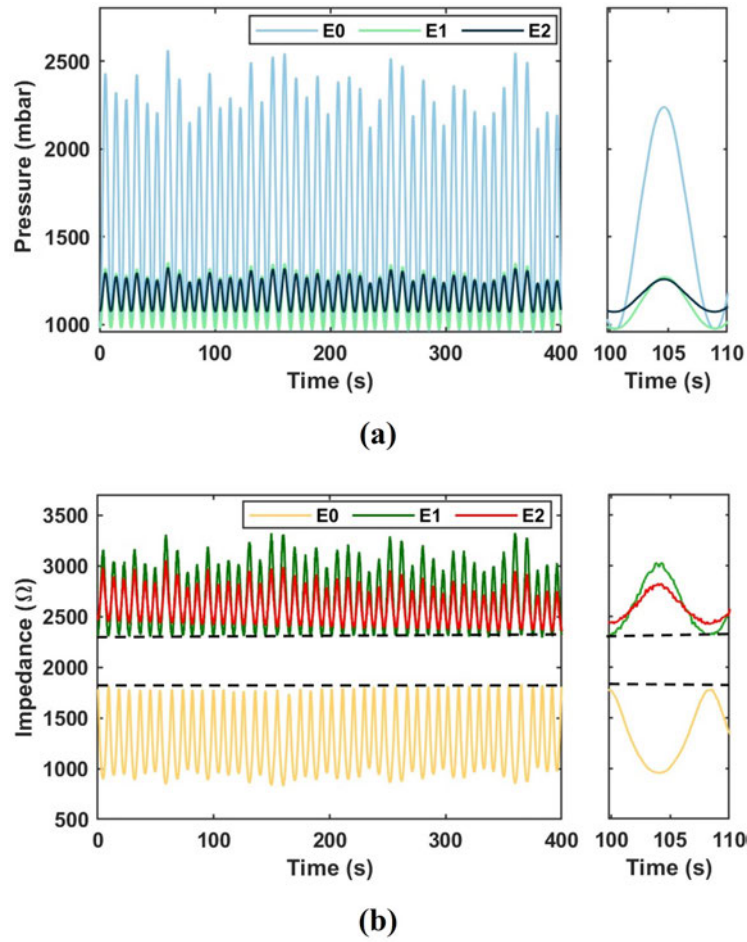


Figure 5.6: *Ex vivo* pressure and EIT raw data with corresponding zoomed views: (a) Random pressure signals with E0 being active, (b) Resultant EIT measurement of Channel 5 per chamber.

clearly indicates the onset of X-ray interference in the EM data. In contrast, the EIT measurements, such as channel 5 corresponding to chamber E (Fig. 5.7(c)), remained stable. Even under zoomed inspection, no additional noise or distortion was detected, confirming the robustness of EIT to X-ray interference.

Assessments under MRI, in Fig. 5.8 show that the SCM equipped with the FPC and saline-filled chambers was clearly represented, whereas the SCM without the FPC was not visible. A magnified view of the SCM with the FPC under MRI revealed the electrodes as distinct dark spots distributed along the working chambers.

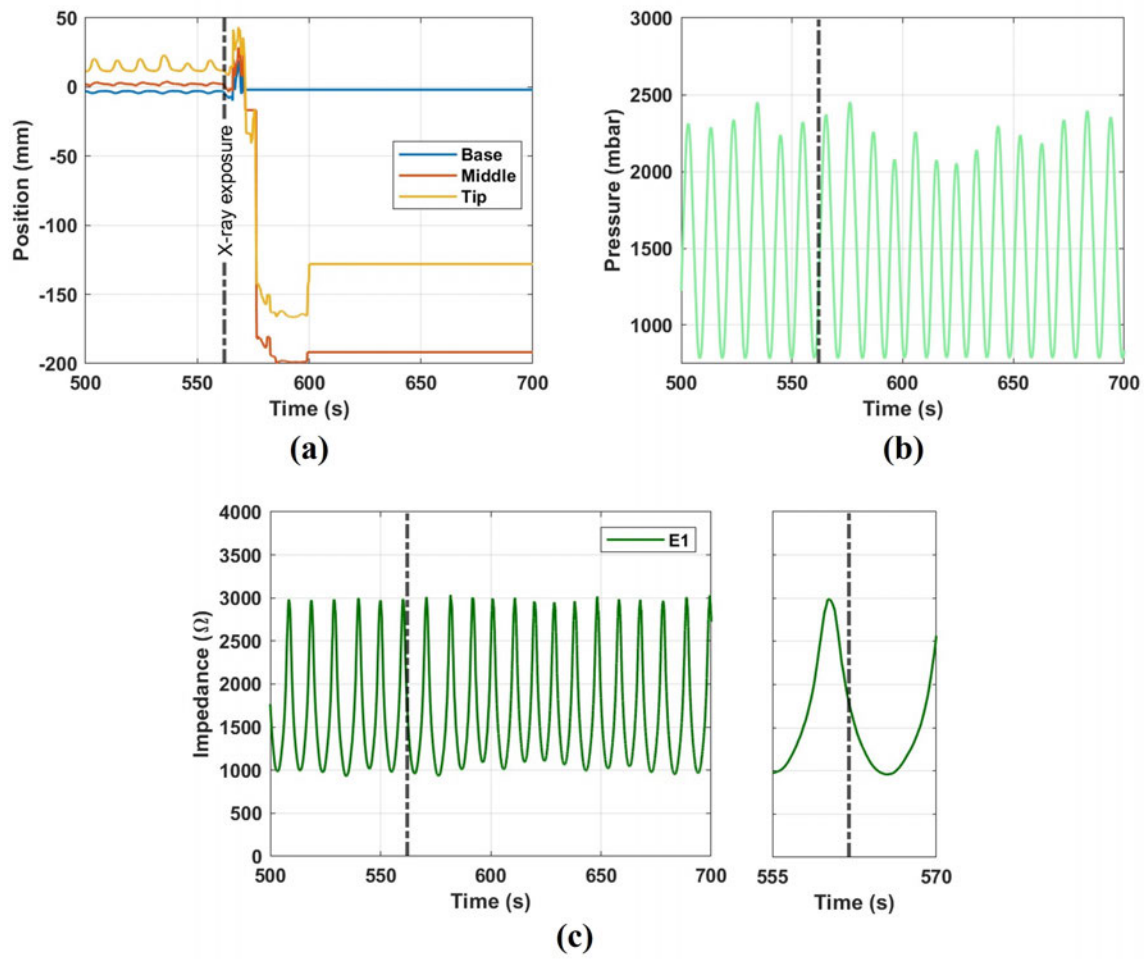


Figure 5.7: EIT performance under the X-ray: (a) Disrupted EM signals in the y-axis after the exposure, (b) Pressure signal applied to chamber E1, (c) Corresponding EIT readings from channel 5 remain unchanged after X-ray interference.

5.3.2 Reconstruction Results

Ex Vivo Tip Localisation

Reconstruction of 7,473 motion instances from the test set using the MLP for predictions can be presented as tip localisation and backbone shape reconstruction. Of the 7,473 instances, 3,028 correspond to E0 activation, 2,091 to E1, and 2,354 to E2 activation. Training on 74,717 motion samples resulted in a training RMSE of 0.1746 mm and a validation RMSE of 0.2465 mm after 14,550 iterations. Relative to free motion (0.05 mm for training and 0.07 mm for validation), these values are 3.492 and 3.521 times higher, respectively, illustrating the increased difficulty of training the model on *ex vivo* data compared to free motion. It should be noted, however, that in free motion the MLP was trained on 540,000 motion samples.

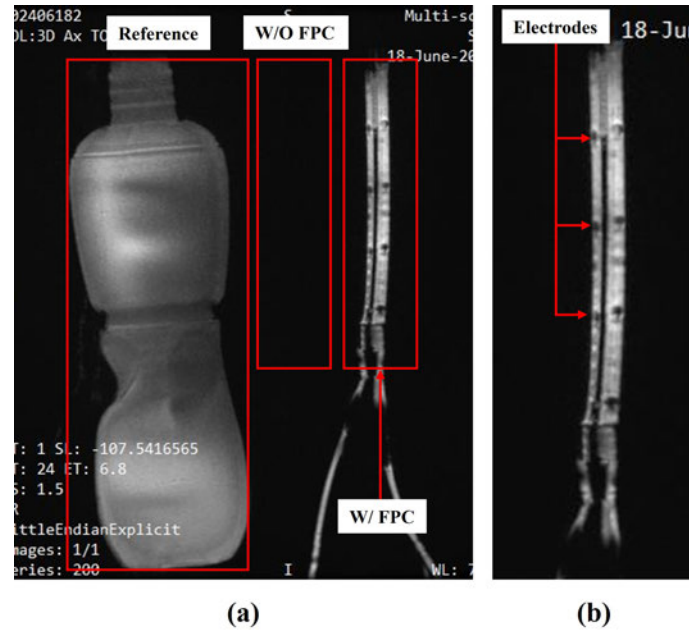


Figure 5.8: EIT examination under MRI: (a) Comparison between visibility of reference, SCM without and with FPC, (b) Distinctive dark spots indicating electrodes.

The transformed predicted tip positions in 3D and their 2D projections are shown in Fig. 5.9, with labels indicating the active chamber for each trajectory. The largest deviation from the ground truth occurs during E0 activation. In the 2D projections, a consistent shift between the predictions and the ground truth is evident for E0 activation. It is important to note that the highest frictional effects were also observed during E0 activation, where the maximum offset between baseline and deflated rest positions occurred. This high error may further be attributed to inadvertent dislocation of the SCM, resulting in baseline shifts between training and test datasets.

A more detailed per-axis analysis in Fig. 5.10 confirms that the largest inaccuracies occur during E0 activation. The highest absolute errors in the selected trajectory along the x-, y-, and z-axes are 5.97, 3.40, and 5.22 mm, respectively. These inaccuracies are most pronounced at local minima and maxima. Furthermore, these regions exhibit higher noise levels associated with variability of the SCM's rest position. Higher errors are caused by the high static friction at the start of inflation and the beginning of deflation. Additionally, error sources include the natural lubrication and mucus present in the colon, which vary with time and location within the tissue. Other uncertain factors include fluctuations in insufflation pressure and complex interactions between the SCM and the haustra.

The RMSE values of tip localisation for different activations, summarised in Table 5.2, show that the largest RMSE values in x, y, and z are 2.86, 1.19, and 1.69 mm, corresponding to E0, E0, and E1 activation, respectively. The average RMSE, calculated according

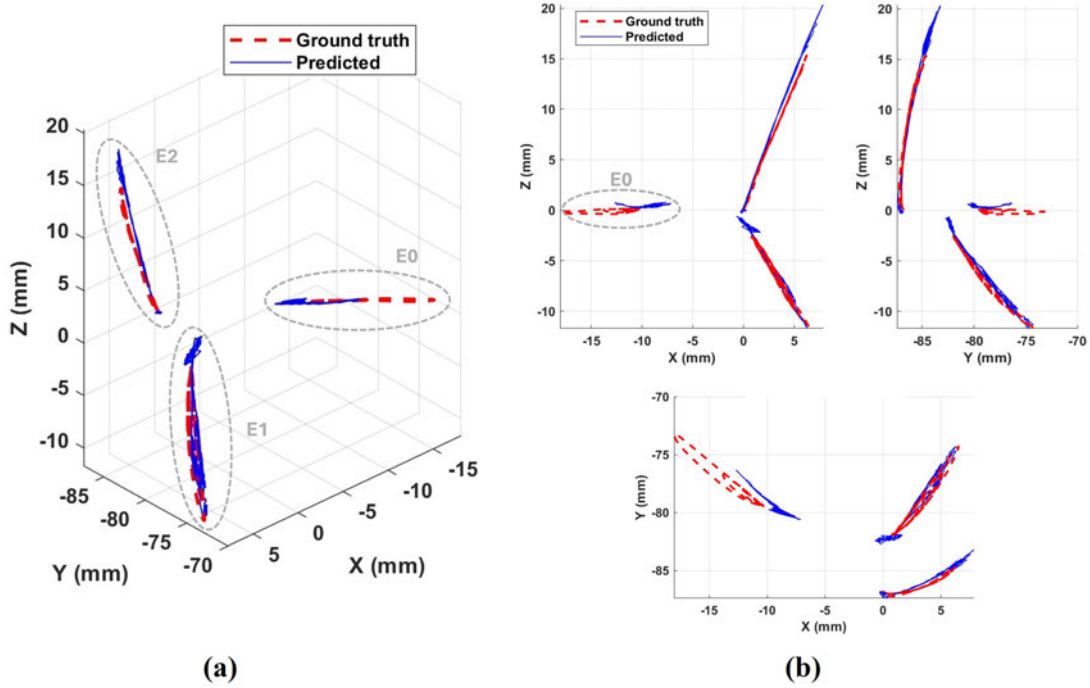


Figure 5.9: Tip localisation performance in *ex vivo*: (a) In 3D space, (b) 2D projections plots. Labels indicate the active chamber for each trajectory.

to Equation 4.15, is 1.82 mm for E0, 1.06 mm for E1, and 0.70 mm for E2, indicating that E2 yields the lowest and E0 the highest localisation errors. The overall error across the test dataset shows that the x-axis is most erroneous, with a relative error of 9.33% of the maximum tip displacement in x. For the y- and z-axes, these relative errors are 5.63% and 3.38%, respectively. By averaging RMSE across the entire test dataset, tip localisation error is 1.34 mm, equivalent to 1.26% of the active length of the SCM. Compared to free motion (0.37 mm), this is 3.62 times higher, reflecting the increased complexity and variability under *ex vivo* conditions.

Table 5.2: RMSE $\pm$ SD of tip localisation error per axis for each activation trajectory.

Activation	e_x (mm)	e_y (mm)	e_z (mm)
E0	2.86 ± 1.06	1.19 ± 0.58	0.57 ± 0.17
E1	0.55 ± 0.55	0.43 ± 0.41	1.69 ± 1.57
E2	0.56 ± 0.42	0.43 ± 0.31	1 ± 0.49
All	1.96 ± 1.69	0.86 ± 0.61	0.89 ± 0.53

Ex Vivo Backbone 3D Shape Reconstruction

To evaluate the sensing approach in terms of shape reconstruction, the localisation accuracy of markers other than the tip was calculated in terms of RMSE and SD, as in

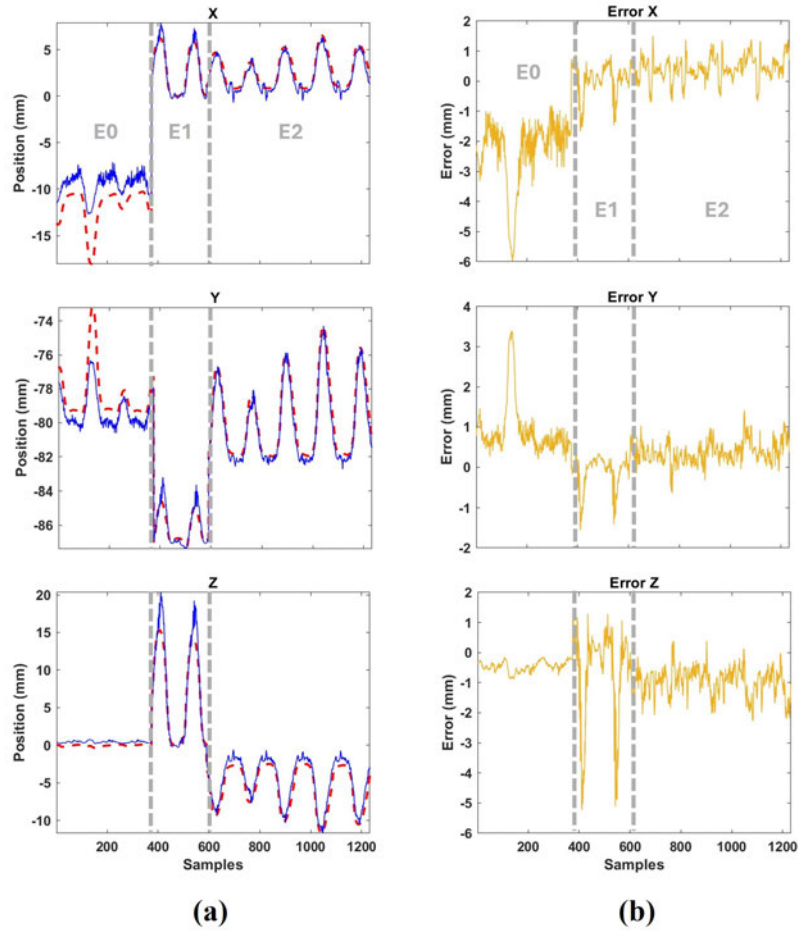


Figure 5.10: Tip localisation performance by axis under *ex vivo* conditions: (a) Prediction versus ground truth, (b) Estimation error. Partitions indicate the trajectory corresponding to each chamber activation.

Table 5.3, per axis and over the entire test dataset. As expected by the correspondingly smaller workspace, the errors are lower for the middle EM sensor compared to the tip. However, as with the tip, errors are greater in the x-axis and smaller in the y-axis. The predicted backbone shape was reconstructed by interpolating 100 points between the base and tip sensors.

Table 5.3: Estimation accuracy (RMSE $\pm$ SD) at different EM sensors' locations over the entire test dataset.

Marker location	e_x (mm)	e_y (mm)	e_z (mm)
Middle	0.95 ± 0.82	0.46 ± 0.28	0.67 ± 0.37
Tip	1.96 ± 1.69	0.86 ± 0.61	0.89 ± 0.53

The fitted reconstructions in the zoomed frames of Fig. 5.11 highlight that inaccuracies

are most pronounced at the tip. Furthermore, the reconstructed shapes exhibit greater nonlinearity compared to those in free motion (Fig. 4.31). This nonlinearity suggests non-uniform deformations induced by variable contact friction and dynamic tissue interactions, thereby highlighting the limitations of PCC models used in the literature to capture backbone shapes of soft continuum robots under *ex vivo* conditions.

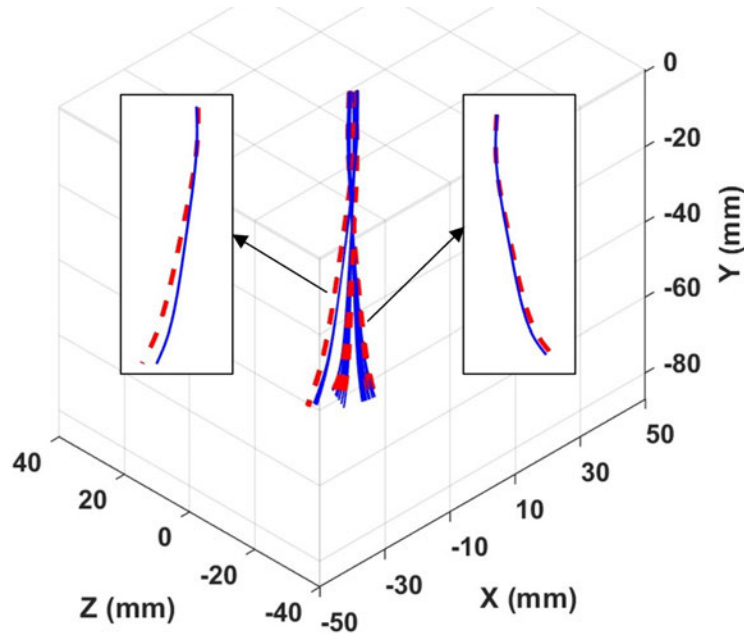


Figure 5.11: 3D shape reconstruction results in *ex vivo*.

To elaborate further on the error progression along the SCM, the RMSE and SD of all 100 interpolated curves are plotted as solid lines and filled areas per coordinate axis in Fig. 5.12(a). For comparison, the equivalent plots from the free motion multi-chamber pressurisation experiment are presented in Fig. 5.12(b). In both domains, the mean error and SD increase from base to tip. However, in *ex vivo* conditions, the error at the base EM sensor is larger than that at the optical markers in free motion. This discrepancy can be attributed to the high tip force generated by local friction, which causes the SCM base to shift, as it is not constrained in the same manner as during free motion experiments.

To quantitatively compare error progression between the two domains, MAE were calculated across the 100 interpolated points. For free motion, multi-chamber pressurisation was selected as it yielded the worst-case error. The results, listed in Table 5.4, show that MAE in *ex vivo* are three to five times higher across all coordinate axes compared to free motion.

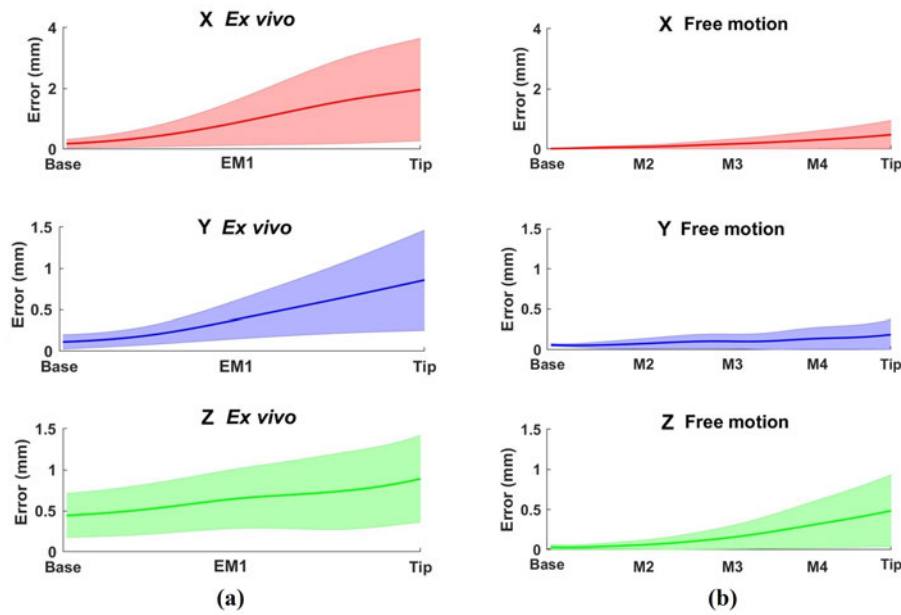


Figure 5.12: Progression of **RMSE** (solid line) and **SD** (filled area) of shape reconstruction from base to the tip between domains: (a) *Ex vivo*, (b) Free motion multi-chamber pressurisation. EM1 denotes the middle **EM** sensor, and M2-M4 label OptiTrack markers in the free motion case.

Table 5.4: Comparison of mean shape reconstruction error between *ex vivo* and free motion multi-chamber pressurisation.

Domain	e_x (mm)	e_y (mm)	e_z (mm)
<i>Ex vivo</i>	0.9668	0.4175	0.6352
Free motion	0.1817	0.1034	0.1799

In vivo Reconstructions

The five configurations (A–E) recorded during the withdrawal procedure of the **SCM**, illustrated in Fig. 5.13, represent the **2D** backbone shapes in X-ray images. After insertion, configuration A was first recorded, followed by sequential withdrawal until configuration E. From A to E, the bending angle of the **SCM** progressively decreases, with A exhibiting the highest bending angle and E the lowest. The backbone was tracked using the camera guidewires, with the tip marked in Fig. 5.13.

During insertion, the surgeon encountered restrictions advancing the **SCM**, since in its non-pressurised state it is highly compliant and susceptible to external bending forces. Despite the use of lubricant gel, substantial axial resistance present at the anus required the surgeon to rotate the **SCM** while simultaneously pushing it forward. After passing through the anus, the onboard camera view became blurred and obstructed due to insuf-

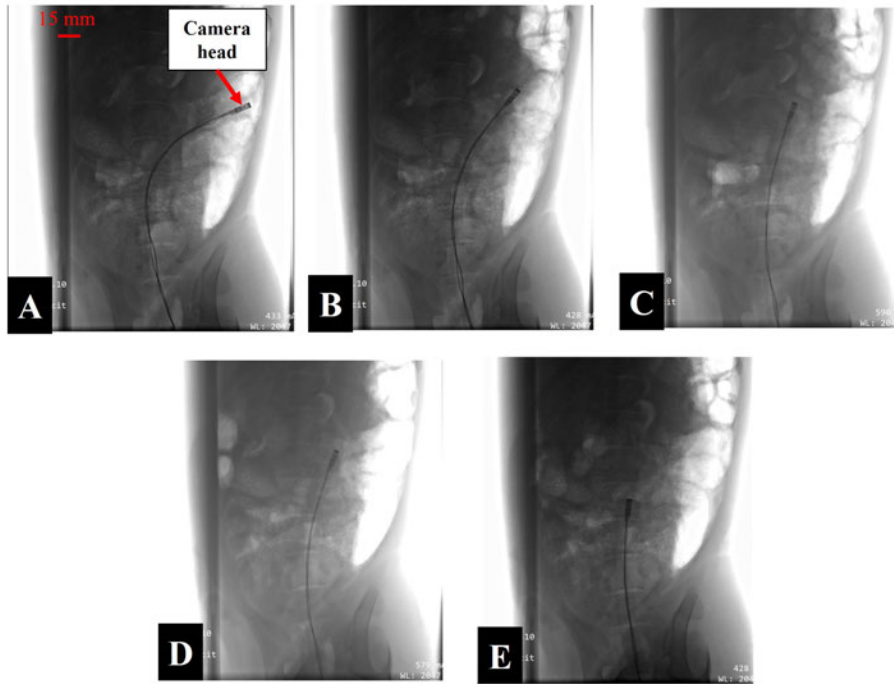


Figure 5.13: X-ray images of the configurations recorded during *in vivo*.

ficient irrigation. Consequently, a conventional endoscope was introduced to improve lumen visualisation and provide insufflation. The surgeon's rotations of the **SCM** during advancement were recorded by the **IMU** mounted on the overtube, as shown in Fig. 5.14(c). However, insufflation could not fully overcome the compressive and axial forces exerted by the colon on the **SCM**, making it difficult to steer the tip towards the tract centre by increasing the pressure in the chambers. This demonstrated the insufficient stiffness of the current prototype for effective active steering in the animal model and overcoming colonic contraction forces. Comparison between the porcine colon in *in vivo* and *ex vivo* experiments clearly indicates that the *in vivo* colon is considerably narrower. This difference introduces a domain mismatch between the collected datasets. Moreover, the larger diameter of the **SCM** compared to conventional colonoscopes (12 mm) results in constant contact with the colonic wall, making active manipulation more challenging.

Fig. 5.14(a) shows the pressurisation of chamber E1 from configurations B to D, increasing from 1023 mbar to 1811 mbar. However, despite the input pressure, no tip deflection was visible in the camera images. The chamber pressure was maintained constant between D and E. Similarly, chambers E0 and E2 exhibited an increase in pressure from B to D (158 and 106 mbar, respectively), but from D to E the pressure remained unchanged.

The corresponding channel 5 **EIT** measurements in Fig. 5.14(b) demonstrate unforeseen behaviour in E1 given the pressure input. An impedance reduction is expected in E1 as

it inflates; however, the opposite behaviour was observed. This indicates that the cross-sectional area of E1 decreased from A to E despite inflation. To assess whether E1 is under tension or compression and to determine its relation to the bending plane in Fig. 5.13, an interval of constant pressure (A–B) is examined while disregarding pressure increases. Between A and B, the bending angle decreases (Fig. 5.13), while the impedance of E1 increases, signifying a reduced cross-section. As E1 is not actively pressurised, this trend suggests that it may experience greater compression at A than at B. Compression causes an unpressurised chamber to expand, but as the SCM straightens, the chamber shrinks and, owing to its elasticity, tends to return to its initial state, thereby increasing impedance. From B to D, a similar shrinkage occurs; however, the simultaneous rise in active pressure expands the chamber, partially counterbalancing the effect.

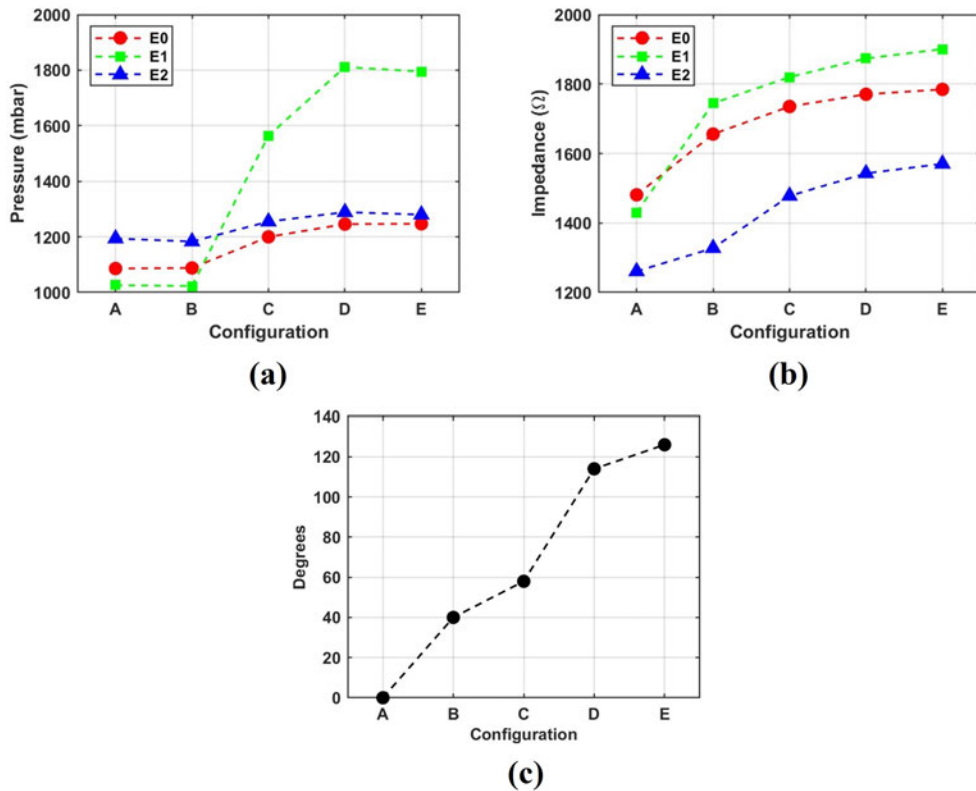


Figure 5.14: Measurements associated with *in vivo* configurations: (a) Pressure, (b) EIT, (c) IMU measured relative to configuration A.

Inspection of the other chambers over the same interval (A–B) also revealed increases in impedance, contradicting the expected opposing trends. Such simultaneous impedance increases across all three chambers were not observed previously, except in cases of axial elongation. This observation likely arises from complex deformations of the SCM during withdrawal. Although the IMU data (Fig. 5.14(c)) provide roll angles, it is uncertain whether relative rotation occurred between the steerable segment and the overtube. Twist-

ing of the chambers during insertion and retraction is therefore plausible. Such twisting, combined with possible longitudinal stretching of the **SCM** during withdrawal, could reduce cross-sectional area in all chambers, hence increasing impedance. Additionally, non-uniform colonic compression must be considered, as local variations in lumen forces may impose asymmetric deformations.

Due to the limited *in vivo* dataset, separate training of the data-driven model was not feasible. Instead, pressure and **EIT** readings from the five configurations were used as a test dataset to evaluate whether the model could reconstruct shape despite substantial distributional mismatch between *ex vivo* and *in vivo* conditions. This evaluation is a preliminary step towards transfer learning: training on a large *ex vivo* dataset, then fine-tuning and testing with the smaller, more complex *in vivo* dataset. Predicted shapes were compared with corresponding X-ray images.

The predicted curves in Fig. 5.15 exhibit discrepancies from the ground truth at various locations along the backbone, primarily arising from domain differences and the relatively passive navigation of the **SCM**. However, the model was able to capture, to some extent, the global trend of decreasing bending angle from A to E: predicted curve A exhibits the greatest deformation, whereas curve E shows the least, agreeing with their ground truth counterparts in Fig. 5.13. These differences were most evident at the tip, although the overall accuracy remains limited. The predicted shapes also exhibit higher-order nonlinearities compared to *ex vivo* results, likely due to the irregular impedance relationships observed *in vivo*, where all chambers showed coinciding increases in impedance.

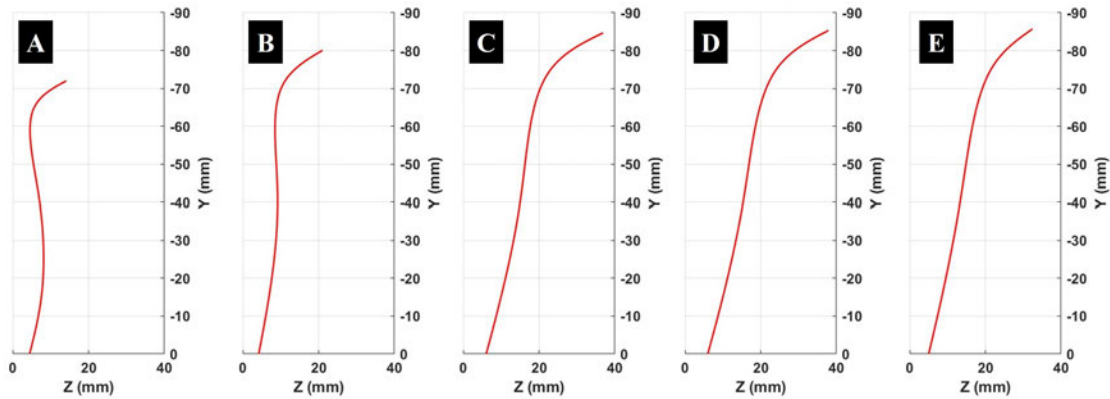


Figure 5.15: Predicted *in vivo* curves by the **MLP** trained on *ex vivo* dataset.

For clearer comparison, predicted curves A and E were superimposed onto their corresponding X-ray images in Fig. 5.16. After scaling and rotation alignment, **2D** reconstruction errors were quantified. Both ground-truth and predicted curves were segmented, interpolated, and compared pointwise, with Euclidean distances calculated in pixels and

converted to millimetres using image resolution. Reconstruction error profiles for configurations A and E are plotted in Fig. 5.16(a, b). Mean errors were 51.03 mm for A and 5.73 mm for E, with maximum errors of 125.49 mm and 12.1 mm, respectively. These results indicate closer backbone reconstruction in configuration E than in configuration A.

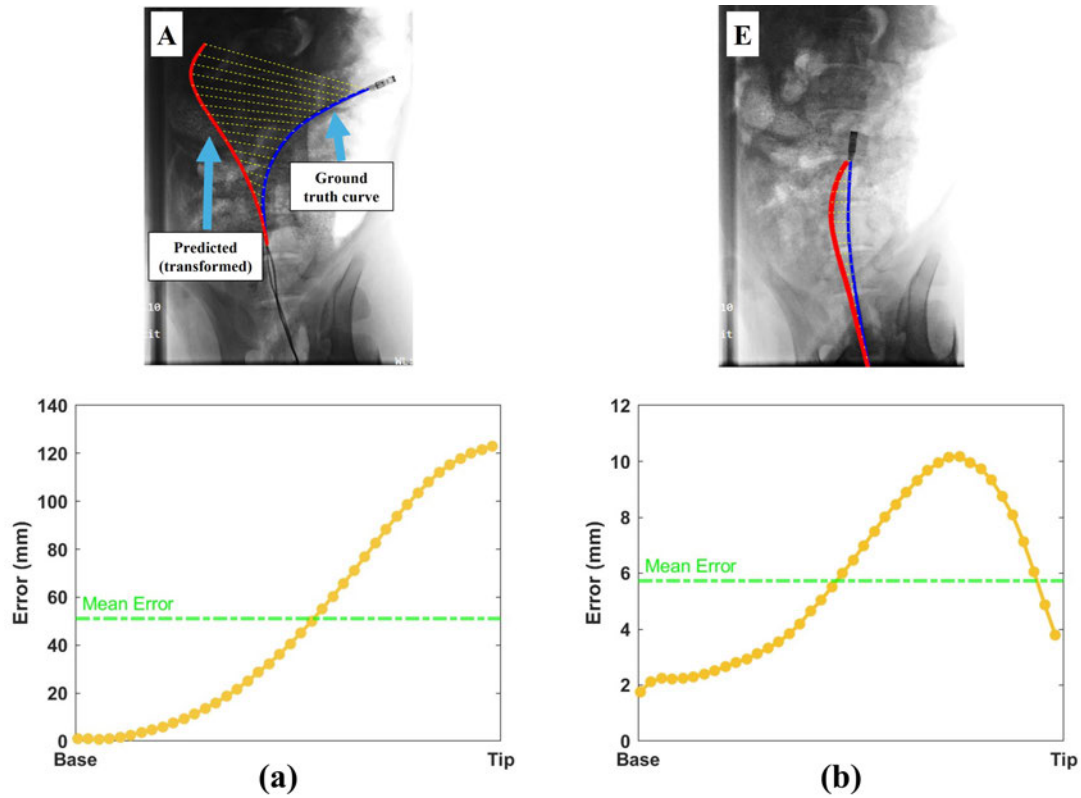


Figure 5.16: Segmenting predicted and actual curves and quantifying the reconstruction performance in *in vivo*: (a) Configuration A and (b) E.

5.4 Discussion

In this chapter, the EIT method achieved an average RMSE of 1.34 mm, equivalent to 1.26% relative to the endoscope length in *ex vivo* planar motion, and a mean error ranging from 5.73 mm to 51.03 mm during *in vivo* trials. In the literature, for pre-clinical assessment of shape-sensing methods, limited studies report quantitative evaluation of shape-sensing performance. Despite their different validation methods, such as the *ex vivo* setup employed or the particular endoscopic application, several results on shape reconstruction and tip localisation can be observed. These results provide context and intuition regarding the shape reconstruction errors obtained in this chapter.

In [224], a relative reconstruction error of 1.15% was reported using FBGs in a fluid-driven, flexible catheter, validated under simplified concentrated loading conditions. In [250], with EMT and an *ex vivo* setup consisting of a 3D-printed phantom of a stomach and pancreatic necroses, an accuracy range of 2.6–6.5 mm was achieved in reaching phantom targets. In [253], vision-based tracking of a rigid surgical instrument was investigated using CNNs, achieving a mean tracking error of 3 mm and a maximum of 5 mm in 3D space for a validation volume of $60 \times 90 \times 80$ mm in XYZ space.

In [251], validation on a silicone aortic arch phantom and a robotic catheter yielded a mean shape reconstruction error between 1.52 and 5.98 mm using intraoperative imaging. A phantom study was conducted in [254] to evaluate the shape-sensing accuracy of an optical method integrated into a catheter. This method was fused with image-guided navigation, and the commercially available fibre-optic real-shape sensing system was utilised. The phantom consisted of a 3D silicone model of an abdominal aorta, and the ground truth was acquired using X-ray imaging. The results indicated a shape reconstruction error of 0.69 mm.

For *in vivo* validation, [255] reported a conventional colonoscope integrated with a four-channel FBG system, each channel comprising five sensors. The sensing net covered 900 mm of colonoscope length. Trials were conducted on an anaesthetised swine, and reconstruction accuracy was evaluated at different nodes along the instrument. They reported an absolute error of 3.77 mm at the starting node, increasing to 57.29 mm near the tip of the colonoscope.

Comparing *ex vivo* and free-motion data, beyond their corresponding shape reconstruction accuracies, an extreme impedance-to-displacement behaviour can be observed in *ex vivo*. Fig. 5.17(a) compares impedance-to-displacement trends in free-motion and *ex vivo* conditions. Initially, *ex vivo* impedance increases without displacement due to static friction, thereby creating a dead zone. A similar phenomenon occurs in the free-motion curve, but

with a substantially smaller dead zone. Beyond 10 mm displacement, the slope reduces in *ex vivo*, indicating less cross-sectional change during continued deflection. Both cases demonstrate decreasing impedance-to-displacement ratios due to the nonlinear elasticity of silicone [236], agreeing with the analysis in Fig. 4.19. To illustrate the dependency of EIT behaviour on silicone elasticity, Fig. 5.17(b) shows that pressure increases at a reduced rate beyond a specific tip displacement threshold, as evidenced by the derivative of the pressure signal with respect to displacement. Given the direct correlation between pressure and the internal volume of the SCM, the volume signal follows a similar trend. Since impedance inversely correlates with volume, EIT values decrease as displacement increases, following a mirrored trend of the volume signal.

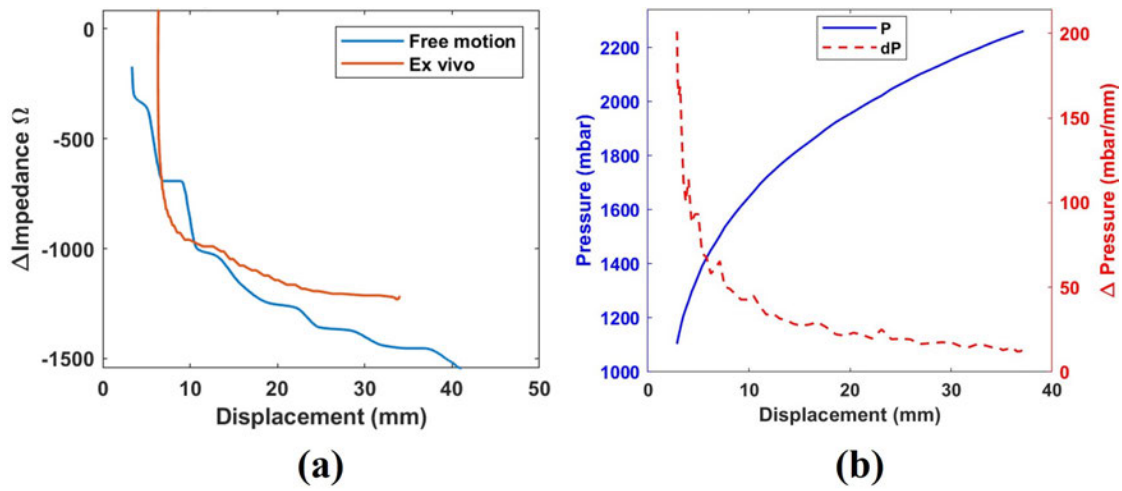


Figure 5.17: EIT values comparison between *ex vivo* and free motion: (a) Impedance vs displacement between two domains, (b) Nonlinear characteristics of the SCM shown by pressure variations against displacement.

To improve estimation accuracies in *ex vivo*, acquiring additional datasets is recommended. Reduced number of data points is one of the factors contributing to the higher inaccuracies in shape reconstruction under *ex vivo* compared to free-motion conditions. Even during training, the RMSE was higher in *ex vivo* compared to free motion (0.1746 against 0.05 mm). Considering that complex environmental factors are more prominent in *ex vivo*, such as the larger dead zone described earlier, increasing the size of the training dataset is strongly emphasised. Additionally, incorporating more EM sensors along the SCM could enhance fitting performance and yield reconstructions that more closely match the actual backbone shape.

In the *ex vivo* setup, colon tissue was mounted on a tray allowing only 2D SCM deflection, which limited data collection to single-chamber pressurisation. Although the data were transformed into 3D using IMU measurements, the setup restricted multi-chamber

actuation. Structuring the colon into anatomically realistic geometries in future studies may yield more generalisable data prior to progressing to *in vivo* trials.

Although sufficient data were obtained in *ex vivo*, clinical settings rarely allow for extensive data collection. High-fidelity simulations replicating mechanical and geometrical properties [256, 202] may serve as an alternative. Sim-to-real approaches, such as [Variational Autoencoders \(VAEs\)](#) [257], enable learning of a latent space shared between simulation and reality, thereby facilitating data augmentation. This method may bridge colon simulators with *ex vivo* data and support generalisation to *in vivo* cases.

While the proposed continuum manipulator was capable of deforming inside colon tissue during *ex vivo* experiments, it lacked sufficient stiffness to actively steer towards the colon centre during *in vivo* trials. This stiffness limitation can be addressed in future work by introducing stiffer materials or by increasing the pressure intake capacity of the current system through greater wall thickness or reinforced sealing of the [SCM](#). For instance, in [48], the outer diameter was set to 17.5 mm, and by reducing the working chamber diameter to 2 mm, greater wall thickness was achieved, enabling the instrument to traverse the colon through self-propulsion. It should be noted that [EIT](#) performance is associated with elasticity as shown in Fig. 5.17(b), hence altering the material necessitates a flexible training strategy. Alternatively, improved steering performance of the soft manipulator could be achieved by further reducing its diameter. A smaller profile would increase the available free space within the target environment and reduce the compressive forces exerted by the surrounding colonic walls. However, such miniaturisation would necessitate further optimisation of both the working channel and the instrument channels to preserve functionality and mechanical integrity.

Assuming stiffness enhancement of the [SCM](#) is achieved and active steering is pursued, the next critical step is to enhance the autonomy of steering and navigation, particularly when the [SCM](#) rotates inside the lumen. As observed, it is not intuitive to decide spontaneously which chamber should be activated, given the [IMU](#) readings of the rolling angle and the rotated endoscopic view. In traditional colonoscopy, this decision relies on the surgeon's expertise, developed through advanced spatial awareness and intensive hand-eye coordination, enabling real-time response to shaft rotation [258]. In future studies, automation of this process using [IMU](#) and camera input is recommended [259].

Autonomous robot-assisted navigation is likely to be essential to validate shape-sensing methods during continuous navigation and deformation of the soft colonoscope, rather than relying on a few discrete configurations. In this chapter, due to the limitations imposed by X-ray radiation, it was impractical and inadvisable to control the [SCM](#) while

the X-ray machine was in operation for capturing ground-truth images. A robot-assisted approach would overcome this limitation and allow the collection of a broader *in vivo* dataset for training and validation purposes. Alternatively, integrating computer vision techniques with DNN-based frameworks capable of autonomous lumen centre detection through depth estimation [260, 261] could further enable autonomous navigation.

A comparison of the data acquired during the *ex vivo* and *in vivo* trials, together with the corresponding reconstruction results, highlights several factors that contribute to the observed discrepancies in accuracy across these two domains. Specifically, comparison of Fig. 5.6(b) and Fig. 5.14(b) reveals a clear mismatch in the baseline EIT values between the *ex vivo* and *in vivo* conditions. This baseline shift poses a significant challenge for the MLP network during *in vivo* shape prediction, as the test data distribution (*in vivo*) does not align with the training distribution (*ex vivo*), leading to the inaccuracies reported in the results section of this chapter. Several factors may contribute to this baseline mismatch, including fabrication variability, differences in saline temperature, and slight variations in solution concentration. To mitigate these effects, future studies should collect training data under a broader range of operating conditions, for example across varying temperatures and electrolyte concentrations, thereby enriching the training dataset and improving model generalisation. In addition, incorporating multi-modal sensing to calibrate EIT outputs against these environmental factors could help decouple their influence from deformation-induced impedance changes. It should also be noted that contact-induced effects may further contribute to baseline drift. As discussed in Chapter 3, the use of EIS offers a potential means to detect and compensate for such contact-related variations, enabling contact-aware calibration strategies.

The discrepancies between the predicted *in vivo* curves and the corresponding ground truth primarily arise from fundamental differences between the *ex vivo* and *in vivo* datasets. Beyond baseline shifts in EIT measurements, notable differences exist in the multi-modal sensing outputs. In the *ex vivo* dataset (Fig. 5.6), an inverse relationship is observed between pressure and EIT output in the active chamber. In contrast, the *in vivo* results (Fig. 5.14(a)) demonstrate a direct relationship between these variables. As discussed in the “*in vivo* reconstruction” section, this inversion can be attributed to SCM elongation or torsional effects induced by interaction with surrounding tissue. Moreover, Fig. 5.14(a, b) shows variations in EIT measurements between configurations (A, B) and (D, E), despite constant internal pressure. This behaviour arises from passive deformation of the SCM during insertion and withdrawal, which introduces externally induced strain independent of actuation pressure. A further discrepancy lies in the proportional relationship between pressure and EIT variation. In the *ex vivo* case, an increase of 1500 mbar results

in approximately $900\ \Omega$ change in EIT, whereas in the *in vivo* dataset (e.g., between configurations B and D), an 800 mbar increase corresponds to only $150\ \Omega$. These inconsistencies suggest that, under *in vivo* conditions, EIT responses are dominated by externally induced deformations rather than purely pressure-driven expansion or possibly leakage, thereby creating a domain shift that degrades the performance of the trained MLP. To mitigate this domain gap, several strategies may be adopted. First, improving dataset generalisation by incorporating training data from a passively deformed SCM, where externally induced deformations are intentionally applied, would enable the MLP to learn representations invariant to environmental interaction. Second, material modification of the SCM to increase structural resilience against external tissue forces may reduce passive deformation effects, thereby narrowing the discrepancy between pressure-driven and environment-driven responses. Collectively, these strategies may improve cross-domain robustness and reduce prediction errors in *in vivo* reconstruction.

To reconstruct shape more accurately with respect to the global frame, it is crucial to measure yaw and pitch alongside roll to determine the orientation of the SCM throughout the trial. This would facilitate the transformation of the reconstructed shape into the actual shape, as shown in Fig. 5.16. However, as stated in the results section, verifying that the SCM does not twist relative to the overtube is challenging. Thus, measuring overtube movement does not necessarily reflect the motion of the steerable segment. Embedding a soft and stretchable on-board IMU into the SCM would therefore be beneficial [262].

During *in vivo* trials, considerable uncertainty exists, including SCM twist, axial elongation, compression forces from the colonic wall, and temperature variations. Each of these factors can challenge the data-driven model and limit its generalisation. It is advantageous to assess the impact of these factors in the laboratory using realistic phantoms before proceeding to animal and clinical trials. With advancements in sim-to-real methods, it may also be beneficial to examine the approach under virtually simulated *in vivo* conditions [263].

Chapter 6

Conclusion

This thesis investigated the application of an EIT-based sensing methodology for SFAs. Throughout the work, the effectiveness and accuracy of the proposed approach were evaluated in conjunction with data-driven modelling techniques. The validation process began with simulation studies and was subsequently extended to experimental and clinical settings. Quantitative results from each stage demonstrated that the proposed method offers comparable, and in some cases superior, performance to existing approaches reported in the literature.

This chapter summarises the key contributions of each chapter and presents the main conclusions drawn from the conducted studies. Finally, the limitations of the current work and potential directions for future research are discussed.

Chapter 1

This chapter discussed the role of proprioception and exteroception in continuum medical instruments, focusing on endoscopy. It reviewed optical, resistive-capacitive, IMU-based, and actuation-sensing methods for state estimation and their applications in twist detection and force sensing.

The chapter started with emphasising the significance of moving towards novel robotics colonoscopy platforms. The drawbacks of traditional colonoscopes have prompted the creation of robotic endoscopy systems that overcome key challenges by offering more stable visualisation, easier training, and improved control of instruments. Steering rather than pushing the device also helps to reduce the occurrence of looping. Moreover, robotic colonoscopy provides superior awareness of the device's configuration and position, where precise shape sensing and tip localisation contribute to safer, more controlled, and less

painful operations.

The evolution of continuum and soft robotic designs further enhances flexibility and safety, enabling smoother movement through the gastrointestinal tract and paving the way for more comfortable, efficient, and autonomous colonoscopy procedures. In the chapter, the main challenges related to proprioception and controllability in soft robotic systems were discussed. It was highlighted that the nonlinear behaviour of soft materials complicated accurate modelling and closed-loop control, hence impeding autonomous navigation. The limitations of conventional sensors highlighted the need for compliant sensory feedback systems capable of integrating shape and force sensing to improve safety and patient comfort in surgical applications.

The literature review outlined data-driven and model-based approaches, including PCC and Cosserat models, alongside Bézier and Frenet–Serret representations. Finally, recent advances in both model-based and data-driven exteroception methods were reviewed, with a focus on embedded sensing technologies.

This chapter guided the aim of this thesis towards the development of a sensing methodology that satisfies the essential requirements for reliable proprioception in medical applications, such as cost-effectiveness, miniaturisation, low profile, electromagnetic immunity, scalability, and both biological and MRI compatibility. These characteristics were further realised through the development of the proposed EIT-based sensing approach.

Chapter 2

This chapter presented preliminary investigations of EIT as a shape sensing technique for soft SCMs, assessing its feasibility and reliability through both simulation and experimental studies. The findings demonstrated that EIT provided a compliant, low-cost, low-profile, and scalable method for enabling tip localisation. Simulation studies using EIDORS and MATLAB highlighted the sensitivity of EIT to shape changes and the influence of electrode number and spacing on sensing performance. Experimental validation confirmed correlations between impedance, bending angle, and applied pressure, with impedance decreasing as the SCM deformed or pressurised.

LSTM-based models were trained to map EIT outputs to tip positions, achieving minimum RMSEs of 3.6 mm and 4.6 mm in the y and z directions, with four-channel configurations providing the best accuracy. The results also showed that measurement protocol and electrode configuration strongly affected localisation performance, particularly in regions experiencing greater deformation. Future work was proposed to use more advanced simulation frameworks, such as Abaqus or SOFA, to capture nonlinear material behaviour

and structural variations, enabling further optimisation of EIT for accurate and robust proprioception in soft surgical robots.

This chapter established the groundwork to investigate whether the proposed method is as effective for tactile sensing as it is for tip localisation. Considering that external forces can deform the morphology of the soft robot, many of the sensing modalities utilised for proprioception, as discussed in Chapter 1, may also hold potential for exteroceptive applications.

Chapter 3

In this chapter, the force estimation capability of EIT combined with an LSTM network was investigated and applied to tissue palpation in medical contexts. The approach aimed to differentiate soft and stiffened tissues, representing healthy and pathological states, while preserving the simplicity of the endoscopic system. In the beginning of the chapter, it was found that contact tip force prediction accuracy strongly depended on the EIT recording protocol and contact location with respect to the tip trajectory. The prediction outcomes yielded that the eight-channel protocol at location B (approximately in the middle of the tip trajectory) achieved the best performance (2.11% relative error), and a one-channel protocol at location A (close to the rest position of the SCM) performed the worst (16.95%). The study was limited to Z-axis tip forces and one-degree-of-freedom in-plane bending.

Upon realising the capability of EIT in force prediction, the tactile sensing functionality of EIT was pursued and successfully demonstrated on simulated tissue phantoms. Tissue palpation capability of EIT maintains the manipulator's structural integrity and enables potential in situ tissue assessment.

Future work was proposed to expand force estimation to three dimensions and multiple contact points along the SCM, enabling simultaneous measurement of force direction and magnitude. The development of bio-realistic colon phantoms, replicating physiological features such as segmentation and haustra, was suggested to allow more accurate and generalisable data acquisition for *ex vivo* and *in vivo* applications.

Additionally, optimisation of EIT protocols, electrode configurations, and measurement strategies could further improve accuracy under complex tissue interactions. The integration of these methods with advanced machine learning models, combined with multi-DOF manipulators, would support robust, real-time tactile perception in endoscopic procedures. The chapter also discussed the use of TDM-based EIT with RNN classifiers, noting that while RNN/LSTM outperform CNN/DNN in temporal tasks, further work is

needed to reduce training time and improve real-time applicability.

Chapter 4

The application of **EIT**-based proprioception in **3D** for reconstructing the shape of a hydraulically actuated **SCM** using an **MLP** network was presented in Chapter 4. The method was validated under single- and multi-chamber pressurisation of the **SCM**, achieving an average **RMSE** of 0.37 mm, which demonstrated high precision and confirmed the potential of **EIT** as a feedback source for closed-loop control in soft robotic systems. The results established the feasibility of employing **EIT** for accurate and continuous shape sensing comparable to state-of-the-art approaches.

This chapter further analysed the deformation behaviour of the **FPC** new design under tensile and compressive loading. Under tensile loading, the **FPC** exhibited negligible out-of-plane deformation up to 90° bending, maintaining stable **EIT** signals over 3000 s. Variations in impedance across channels were mainly caused by differences in saline depth and uneven inter-electrode stretching. Under compression, however, some regions of the **FPC** detached from the silicone substrate, leading to noticeable out-of-plane deformation, hence producing greater signal deviations.

Additionally, the electro-mechanical behaviour of the integration of the **FPC** into the three-chamber **SCM** was characterised. It was demonstrated that apart from the active chamber (the pressurised chamber), the other two passive chambers exhibited secondary pressure increases due to mechanical coupling. The **EIT** readings reflected these dynamics: impedance decreased in the active chamber and increased in the passive ones, consistent with chamber expansion and compression, respectively. Further evaluations included signal repeatability, **EIT**-tip position correlations, and hysteresis.

The 3D **SCM** was designed with dimensions consistent with conventional colonoscopes to ensure clinical compatibility. Nonetheless, future optimisation of the fabrication procedure such as chamber geometry, actuator length, and wall thickness could further enhance its manoeuvrability and flexibility in future iterations. For instance, incorporating additional working chambers beyond the current three is expected to improve motion smoothness during multi-chamber actuation. Additionally, the integration of miniaturised camera heads is also anticipated to support further downscaling for broader endoscopic applications, including bronchoscopy. Introducing materials that are compatible and bondable with the **SCM** material for the connecting tubes and **FPC** is recommended, as this would simplify the sealing process and enhance overall flexibility, respectively.

Complementary **FEA**-based simulations of the **FPC** design were suggested to analyse

structural effects and fatigue behaviour under cyclic deformation. Evaluating the FPC's lifespan, including potential cracking or degradation, would inform improvements in sensor durability and reliability. Substituting the polyamide substrate with an elastomer-based alternative could be identified as a promising direction to achieve greater flexibility and durability. Although elastomers might introduce hysteresis.

Further refinement in fabrication methods was also addressed. Replacing flexible PLA moulds with stiffer materials would ensure more uniform chamber dimensions, while automation in moulding and FPC attachment could significantly improve repeatability at the same time. Finally, validating the proposed EIT-based shape-sensing method in *ex vivo* and *in vivo* settings was identified as a critical step toward clinical translation, building upon the promising accuracy and robustness demonstrated in this study.

Chapter 5

This chapter extended the sensing framework, previously validated under free-motion conditions, to clinical environments through *ex vivo* and *in vivo* trials. The objective was to assess the robustness of the EIT-based shape reconstruction method integrated with the MLP network developed in Chapter 4. For the *ex vivo* phase, a porcine colon model was employed to replicate the mechanical characteristics of the anatomical tissue, whereas the *in vivo* experiments were conducted on a 35 kg porcine subject under anaesthesia.

The experimental setup used in the *ex vivo* trial was detailed in this chapter. Positional data were acquired using an EMT system, with three five-DOF EM sensors mounted on the SCM via silicone belts. As visualisation of the endoscope's rolling angle was obstructed, an IMU was additionally affixed proximally to monitor torsional orientation during insertion.

Prior to experimentation, safety evaluations confirmed the compatibility of the EIT system with clinical imaging modalities. MRI testing using a 1.5 T scanner verified the absence of magnetic attraction, projectile motion, or image artefacts induced by the FPC. Similarly, X-ray trials demonstrated no signal distortion during exposure.

During the *in vivo* trials, data acquisition commenced following colon insertion and continued throughout controlled retraction, with X-ray imaging providing 2D projections of the SCM's shape. Due to radiation safety constraints and consequently the limited continuous presence of clinical personnel, data were collected at discrete static configurations. The datasets served as unseen test inputs for the MLP model trained on the *ex vivo* data.

The reconstruction results indicated that EIT system achieved an average RMSE of 1.34 mm

(1.26% of endoscope length) during the *ex vivo* trials. The impedance–displacement relationship exhibited a dead zone under *ex vivo* conditions, attributed to static friction, and a nonlinear decline in sensitivity at larger deflections due to silicone hyperelasticity. These phenomena are considered major contributors to the increased reconstruction error compared with free-motion experiments.

In the *in vivo* setting, the reconstruction error ranged from 5.73 mm to 51.03 mm. Although relatively high, these values are expected given that this was the first animal trial evaluating the proposed sensing system, with several experimental limitations contributing to reduced reconstruction accuracy. For contextual comparison, other reported animal studies in the literature were reviewed to establish a reference for achievable accuracy.

Under *in vivo* conditions, domain mismatch between active (*ex vivo*) and passive (*in vivo*) deformation, as well as factors such as twist, axial elongation, wall compression, and temperature variation, adversely affected model performance. To enhance generalisation, future work should include *ex vivo* data collection under passive loading conditions and employ domain adaptation or sim-to-real strategies to improve robustness and clinical reliability.

Concluding Remarks

Across this thesis, **EIT** demonstrated its capability as a promising approach for perceiving the states of soft continuum robots in both proprioception and exteroception contexts. Its notable accuracy in both planar and spatial domains highlights its potential for integration within feedback systems for closed-loop control architectures. Although the data-driven methods explored in this thesis were primarily based on **RNN** and **ANN** architectures, numerous other network designs could be investigated to further enhance reconstruction accuracy. Even optimisation of the hyperparameters within the existing architectures presented in this work could serve as an effective starting point for improving reconstruction outcomes.

Different variations of **EIT** protocols were adopted throughout various chapters. Although many of these protocols yielded satisfactory estimation results, it was observed in the earlier chapters that the readings varied depending on the number and spatial configuration of the channels. The promising outcomes achieved with **EIT**, even without protocol optimisation, demonstrate the robustness and reliability of the method. Nevertheless, to fully realise the potential of **EIT**, future studies should explore optimised protocols tailored specifically to either proprioceptive or exteroceptive applications. Furthermore, conducting ablation studies to evaluate the influence of individual channels on reconstruction

accuracy while keeping the ML model architecture and hyperparameters constant would provide valuable insights. This ablation process, however, poses challenges, as the complexity and tuning of data-driven models are highly dependent on the feature space size. A strong interdependence exists between the number and type of channels employed and the characteristics of the corresponding data-driven model.

This study unveiled several avenues for future research beyond those outlined in the individual chapter contributions. One of the key advantages of employing EIT is its dual functionality as a hybrid actuation–sensing mechanism; however, it also holds promise as a standalone sensing modality in passive scenarios. Accordingly, EIT could be integrated into a soft fluidic sensing skin, implemented as either a sleeve or a modular add-on. Owing to the scalability of EIT, such a sleeve could be desirably miniaturised and mounted onto existing colonoscope platforms, including cable-driven systems, offering distinct actuation and sensing mechanisms. Similar strategies have been demonstrated using alternative sensing modalities [252]. This approach effectively decouples the influence of actuation from deformation, enabling more isolated and precise shape detection. This modification is essentially effective in reducing the dead zone present in the pressure-tip location correlation. Furthermore, it opens the possibility of retrofitting conventional colonoscopes with advanced sensing capabilities without necessitating a complete redesign.

In vivo trials demonstrate that the SCM may undergo passive deformation in situations where active steering is not feasible, for instance due to insufficient structural stiffness or limitations in accurate tip navigation. Prior to the *in vivo* experiments, all investigations were conducted under conditions of purely active deformation, where shape change was driven exclusively by internal pressurisation. These contrasting scenarios indicate that the sensing and reconstruction framework must operate reliably under both active and passive deformation instances. Furthermore, it was observed that the EIT response during active pressurisation differs from that arising from passive, externally induced deformation. To ensure that the data-driven reconstruction model remains valid across both conditions, the current single-model framework can be replaced with two dedicated machine learning models: one operating when active deformation is detected, and the other reconstructing shape under passive deformation. Each model would be trained on a domain-specific dataset, and their deployment could be preceded by a classifier that determines whether the observed motion is active or passive. Such a modular architecture may yield more efficient training and improved reconstruction accuracy compared with training a single model on a combined dataset comprising both passive and active deformation patterns.

The SCM employed in Chapters 4 and 5 for spatial validation of EIT comprised a single three-chambered segment, offering two DOFs. Nonetheless, EIT can be extended

to multi-segmented **SCMs**, as demonstrated in [264], where serially integrated **SCMs** enhance the dexterity of soft endoscopes. This scalability, however, necessitates an increased sensing capacity, thereby raising the number of required ports for the impedance spectroscopy. Such challenges can be mitigated either by optimising the number of electrodes per working chamber or by employing a spectroscopy with a higher channel count.

Another direction worth pursuing concerns hybrid and simultaneous force–position estimation using **EIT**. In this thesis, the capability of **EIT** was investigated for each of these sensing objectives independently, but not concurrently. One limitation in achieving this integration arises from the intrinsic compliance of soft robots; their deformable nature makes them highly susceptible to external stimuli. Consequently, internal deformations within the working chambers are influenced by both actuation pressure and external forces, making it challenging to decouple these effects. A potential solution, while maintaining adherence to the **EIT** framework, would involve introducing additional channels, preferably positioned near the exterior of the **SCM**, to specifically capture environmental interactions, thereby facilitating the decoupling of exteroception from proprioception.

One of the major challenges associated with data-driven modelling lies in the acquisition of high-quality datasets that meet the requirements for generalisation, accurate labelling, and consistency. In this study, it was determined that the **EIT** response is strongly influenced by the strain–stress behaviour of silicone or other soft composite materials. This dependency complicates the data collection process, particularly given that fabrication techniques for soft medical robots are not yet firmly repeatable. Consequently, in case of replacing the soft robot used for integration, the entire data acquisition process may need to be repeated. A promising alternative would be to explore model-based shape reconstruction using **EIT**, wherein the correlation between **EIT** outputs, the expansion rate of the elastic body, and its final deformed configuration is analytically formulated. This process could be accelerated through established model-based approaches such as the Cosserat rod theory or the Frenet–Serret method.

In the present work, the proposed platform was successfully validated under *ex vivo* conditions, yielding sufficient data to train the neural network model. However, in typical clinical evaluations employing data-driven methods, it is often neither cost- nor time-effective to design and conduct experiments capable of producing comprehensive and generalisable datasets. One potential strategy to mitigate this limitation is the use of high-fidelity simulation environments that accurately reproduce clinical scenarios, incorporating the mechanical properties of the colon, lubrication, and surface friction [256], alongside anatomically accurate colon geometries [202]. Validating such simulation environments can significantly accelerate model training and reduce the experimental burden

associated with clinical data acquisition.

During *in vivo* trials, the endoscopic camera integrated within the SCM failed to adequately visualise the surgical site, primarily due to the absence of an irrigation channel. In future iterations of soft colonoscopic instruments, incorporating an additional channel for irrigation could effectively address this issue. Moreover, the dual functionality of EIT offers the potential to free internal space for integrating supplementary surgical tools such as the *iKnife*, biopsy forceps, or other interventional instruments within the working chambers. In this thesis, only an endoscopic camera was integrated; however, extending this integration to include additional instruments, while investigating their effects on the mechanical properties of the SCM and on EIT readings, would further substantiate the claims regarding the scalability and miniaturisation potential of EIT. Furthermore, the current design parameters of the SCM, as demonstrated in *in vivo* experiments, suggest that reducing its diameter would be advantageous. The shape-sensing approach using EIT supports this miniaturisation, as it is inherently scalable. During *in vivo* trials, the SCM remained in constant contact with the colonic wall, hindering deflection and active navigation. A smaller diameter would create additional clearance within the lumen, thereby reducing contact friction between the instrument and surrounding tissues.

In addition to its demonstrated utility in medical robotics, EIT has significant potential in a range of non-medical soft robotic applications. For example, the use of EIT for continuum tactile perception and deformation sensing has been explored in soft robotic skins and flexible manipulators, enabling high-resolution contact and force estimation, which is particularly relevant for soft fluid-driven grippers handling delicate objects [265]. In locomotion-oriented soft robotics, distributed proprioceptive sensing could enhance state estimation and control of hydraulically actuated continuum robots navigating confined or unstructured environments [266]. Furthermore, advanced tactile and deformation sensing via EIT can support efficient perception and control in assistive or prosthetic robotic hand systems, where miniaturised, embedded sensors are critical for estimating joint configuration and improving dexterity [267]. Across these domains, EIT contributes to enhanced system intelligence by enabling embedded perception that informs control decisions for compliant, adaptive soft robots.

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Appendix A

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Multi-Interface Strain Transfer Modeling for Flexible Endoscope Shape Sensing

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MorphGI: A Self-Propelling Soft Robotic Endoscope Through Morphing Shape

Author: Julius E. Bernth, Guokai Zhang, Dionysios Malas, et al

Publication: Soft Robotics

Publisher: Mary Ann Liebert, Inc.

Date: Aug 1, 2024

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Conference Proceedings: 2024 IEEE 7th International Conference on Soft Robotics (RoboSoft)

Author: Adira Colton

Publisher: IEEE

Date: 14 April 2024

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