

# **Advancing the Development and Integration of Dynamic-Bodied Ionic Solution-Based Soft Sensors for Biomedical Robots**



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## **Declaration**

I hereby declare that, except where specific reference is made to the work of others, the contents of this dissertation are original and have not been submitted in whole or in part for consideration for any other degree or qualification in this, or any other university. This dissertation is my own work and contains nothing which is the outcome of work done in collaboration with others, except as specified in the text and Acknowledgements.

Joanna Elizabeth Jones

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## Abstract

Medical robots are increasingly adopting soft robotic technologies, to minimise the damage left behind by robotic interventions and explore new therapeutic treatments. For long-term implantable robots interacting with tissue, there are a plethora of challenges presented by the in vivo context, including biocompatibility, safety and compactness of design. Soft robots afford many advantages and answers to these contextual requirements, including their inherent compliance and hyperelasticity, providing the means for both increased safety, as well as generating large expansions from compact designs. However, the use of soft robots is limited by their infinite degrees of freedom, state-dependent and nonlinear behaviours, a strong dependence on the external environment, and difficulties in differentiating between stimuli. Soft sensors capable of giving information on the system and environment as a whole, as well as any abnormalities and faults, and, importantly, able to adapt to the changes in both the robot and the environment, will be key in driving, and securing, the future of soft technologies in medical robots.

In this thesis, a variety of sensing technologies are implemented and explored to advance the development of safer and more effective medical robots. Firstly, a soft sensor-actuator, capable of large expansion and differentiating between height and pressure, is introduced as a building block for a variety of demonstrated probe and implant applications. Pressure and resistance sensing were shown to be interchangeable, although best used in combination, and a best estimated resolution of 4 g and 0.07 mm was achieved. The integration of these sensing technologies to distinguish between internal and external stimuli is developed further in the context of tissue lengthening soft robots. A three-columned sensor-actuator is shown to have good structural strength, shrinking by only 12 % under a 500 g load and growing more than 80 % of its unloaded height when preloaded with 250 g. The resistance sensing also proved useful for bending characterisation, but further work is still needed to improve repeatability. Finally, new methodologies, to maximise the information of localisation of stimuli and to tune the sensors to a robot and its context, are presented. The ability to tailor and tune sensing is preliminarily demonstrated as an opportunity to adjust sensing dynamically to the context. Additionally, exploiting the dynamics of fluid-based sensing is shown, allowing single and multiple press events and their locations to be determined.



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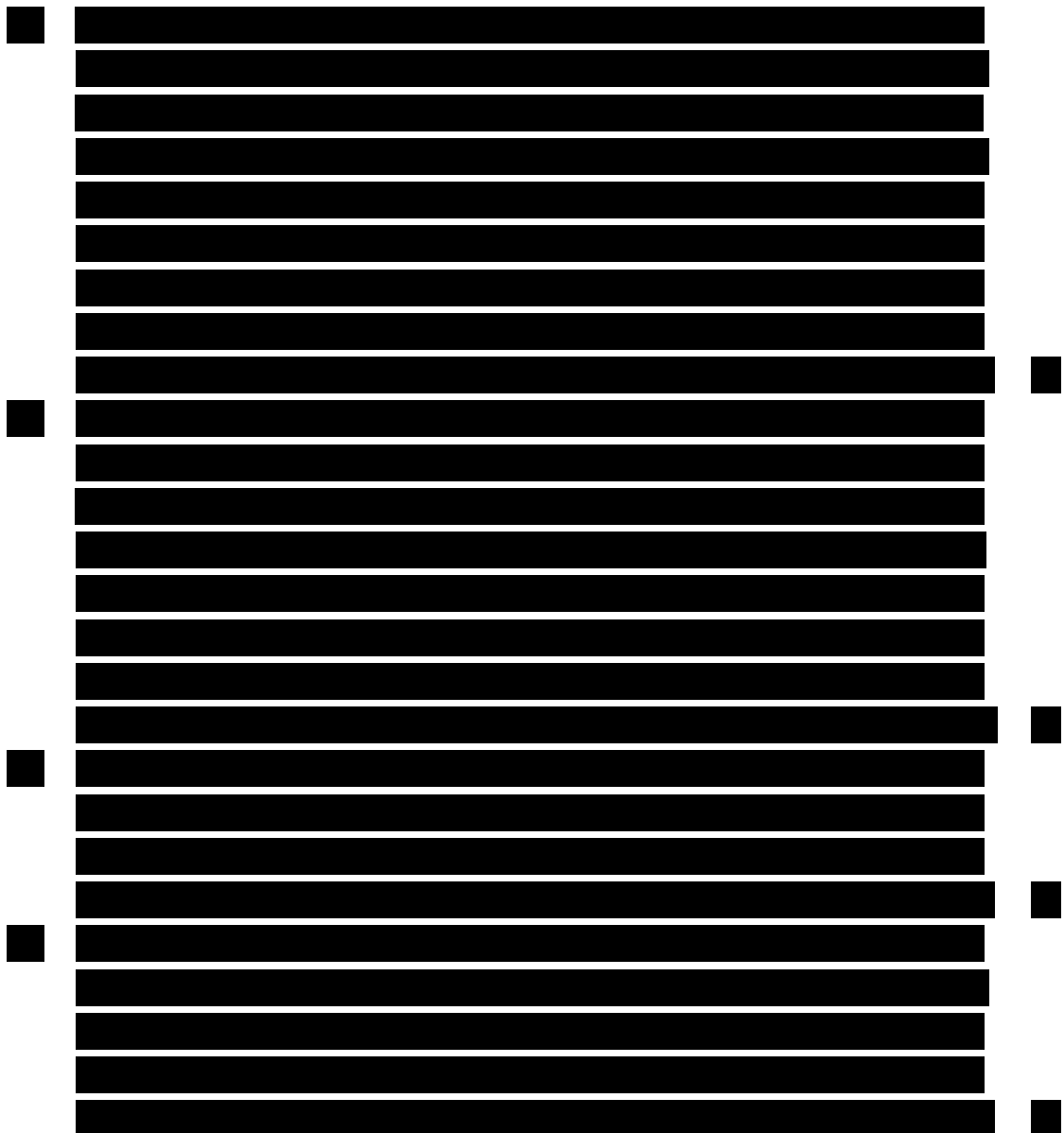
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# Chapter 1

## Introduction

### 1.1 Context

This project is anchored in the context of robotic implants designed to promote growth in tubular organs along the gastro-intestinal tract (GI) to treat conditions, such as long gap oesophageal atresia (LGOA) and short bowel syndrome (SBS). LGOA is a condition where a baby is born with a gap in the oesophagus of 3 cm or more, and SBS is a condition where a patient has less than half of their bowel remaining due to surgical intervention or a congenital defect.

The robotic implants designed in the Sheffield Biomedical Robotics Lab (SBRL) aim to replace the current methods for treatment and allow healthy normal living. For LGOA, the current treatment is the Foker technique, which involves placing sutures on the ends of the oesophagus, and gently pulling on them from the baby's back to slowly grow the tissues. Throughout the procedure, the baby is sedated and overall the procedure carries many risks including the sutures breaking. For SBS, the current treatment involves a variety of different approaches depending on the severity of the condition. For more mild cases, the treatment involves supplementary parenteral nutrition, oral rehydration solutions and medications to increase hormone growth or reduce motility. In more severe cases, surgery to lengthen the remaining small intestine is carried out (serial transverse enteroplasty procedure or Bianchi procedure), and in the most extreme cases intestinal transplant.

Other research groups have been developing solutions for tubular tissue growth and have reached in vivo animal trials. One group with a preliminary study on mechanical enterogenesis lengthening in an infant with SBS [2], has shown the ability to grow human tissue, using a balloon tip catheter and gradually pulling a portion of the ileum with the distal end anchored to the retroperitoneum. The same group has also investigated the use of springs to induce the tissue growth, whilst leaving an intraluminal passage for chyme [146],

including a long-term safety study [145]. The spring-mediated solution shows promise for multiple sections of lengthening without obstructing normal digestion, but has only shown limited growth in animal trials. Another group has been tackling the issue with a telescopic expansion actuator [45][148], allowing larger growth, but still limited to the size of the body cavity given its rigid nature. Both research groups have also noticed damage to the tissue as a result of the lengthening process. One other group has shown the use of shape-memory polymer [55], offering a softer approach, but this has only been tested on rats and is an extraluminal device requiring a more invasive surgical implantation. There is therefore still a gap for new solutions to address the following: using softer approaches to more closely match the tissue to be lengthened and surrounding tissues; ensuring the device can be placed intraluminally using minimally invasive approaches; and embedding sensing capabilities into the devices to allow more targeted and informed decisions.

In SBRL, some notable works include the initial design of a rigid robotic implant [39] and its use in animals trials [40], the development and control of a flexible robot based on this design [9, 4], as well as work on an entirely soft alternative [140, 138] and its testing on *ex vivo* tissue (Fig. 1.1). The stiffness was a limiting factor for the rigid and flexible robots, while the soft implant was limited by size and its primitive sensing, leaving a clear gap for investigation.

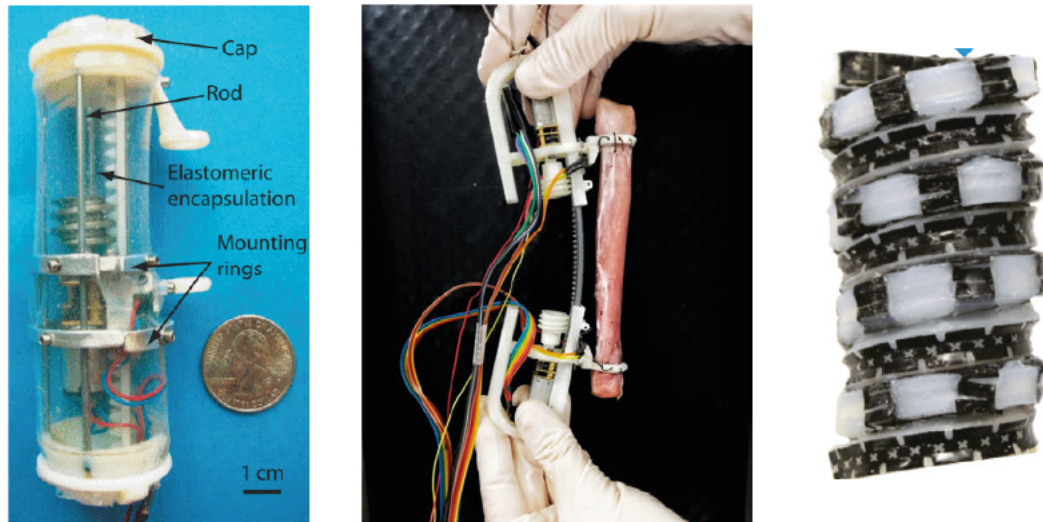


Fig. 1.1 Evolution of implants developed in SBRL. (Left) Rigid robotic implant. Reprinted from [39], ©2014 IEEE. (Middle) Flexible robotic implant. Reprinted from [4], ©2020 IEEE (Right) Soft robotic implant. Reprinted from [138], ©2020 IEEE.

## 1.2 Challenges

With implantable technology being safety critical, the need for sensors to give a more complete picture, in order for the controller to make safe and well-informed decisions, cannot be understated. Making these sensors soft and easily integrated is also essential, so that their presence is as minimally invasive as possible. There has equally been very little work to date, on developing fault-resistant robots that are capable of detecting and localising faults before any potential catastrophic failures, and instead giving time to take preventative action. Hence, this work, which aims to develop soft sensors around resilience and safety, is very much needed.

However, the nature of the project presents challenges to be overcome which are two-fold, those coming from the technology used and the target applications envisaged.

Firstly, the technological challenges that arise from soft robotics:

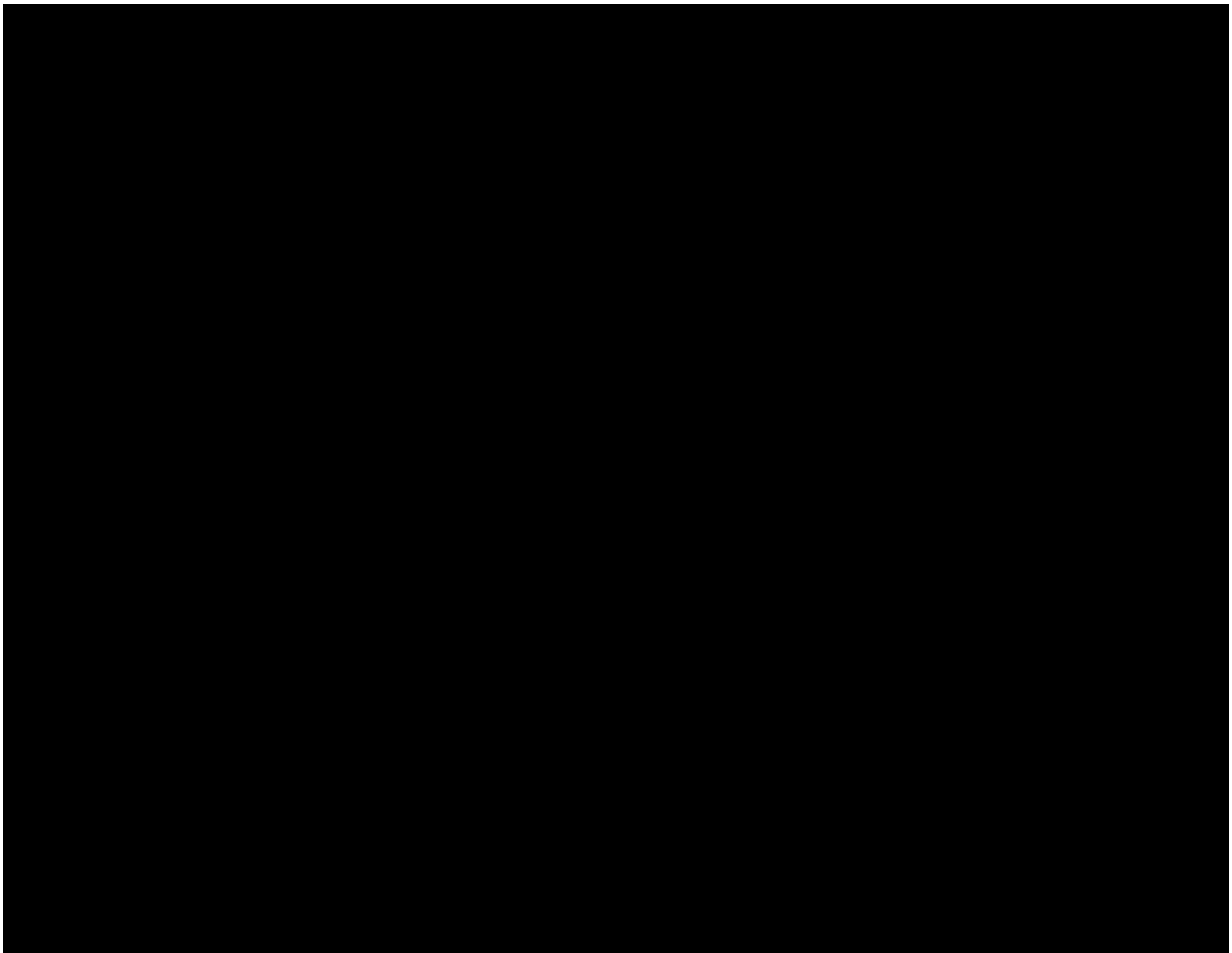
- Infinite degrees of freedom: given the softness of the materials used and their compliance in all directions, controlling the motion in any direction, as well as developing sensors to capture these movements, is more difficult than with conventional rigid components;
- State-dependent and nonlinear behaviours: the use of soft materials, although giving lots of freedom, does come with the added difficulty of state-dependent behaviours (e.g. hysteresis) complicating the controller design;
- High dependence on external environment: with the high compliance and degrees of freedom, soft robots are incredibly susceptible to external forces, making the robot and the environment highly interdependent and making tracking of both essential;
- External stimuli differentiation: given the high dependency on the environment, the need to maximise the sensitivity to specific external stimuli is crucial.

Secondly, working towards *in vivo* applications brings with it its own unique set of challenges:

- Biocompatibility: using materials that are non-toxic and non-harmful to human tissue is a necessity, as well as extending this to minimise the foreign body response;
- Safe and fault-tolerant: given the safety critical aspect of *in vivo* testing, ensuring the overall safety of the system and making it resistant to faults, before potentially harmful behaviours, is paramount;

- Lightweight and compact design: making the system as non-invasive as possible is also essential, as well as making the external hardware lightweight and compact, to ensure that the implant is of minimal hindrance and allows normal daily living.

In order to address the technical and clinical challenges described above, a number of design requirements to be fulfilled have been summarised in Table. 1.1.



## 1.3 Aims and Objectives

### Aims

The aim of this project is to enhance the sensing capabilities, and ultimately the safety, of soft medical technologies, and in particular the previously developed soft implants in SBRL (e.g. Soft Pneumatic Helically Interlayered Actuator (SoPHIA) (Fig. 1.1)). To this end, this project



focuses on the development of safe, biocompatible and information-rich sensors which can be integrated into soft robot bodies, including the investigation of combined sensing and actuation to minimise overall footprint. The sensors are to be designed for *in situ* sensing, with a focus on miniaturising their size, to allow easy integration. Within the context of SBRL's implants, these developed sensors would be used to complement, and/or replace, the already existing *ex situ* pressure, distance and force sensors, and will therefore need to easily integrate into the implant design, and minimise the space and equipment needed. After their development, the sensors will enable the future investigation of closed-loop strategies and control strategies aimed towards fault-tolerance and resilience.

To achieve safety in soft robotic implants, three key sensing areas to target have been identified:

1. Standard sensing: global view of how the robot is acting on and interacting with an environment (interoceptive and exteroceptive sensing).
2. Abnormalities/fault localisation: identifying specific locations of faults or abnormalities (e.g. due to overinflation or interaction with the environment), as well as understanding regions of concerns within the environment interactions, to allow action in the control of the robot to be taken.
3. Sensing adaptation: tuning the sensor to different applications, or to account for system change (e.g. growth), and maximising sensing information using dynamic sensing tuning.

## Objectives

The general hypothesis of this work is that soft sensors can improve the safety and therapeutic effectiveness of implantable robotic devices. Drawn from the clinical and technical requirements provided in the previous section, and to achieve the aforementioned aim, the specific objectives of this thesis are:

- Design, fabricate and validate soft and safe sensor-actuators that can be integrated into an extensive range of soft medical robots
- Demonstrate the sensor-actuators in several clinically-relevant demonstrations, working within size constraints of gastrointestinal tissues and achieving required force and lengthening sensing capabilities, and the minimum force to stimulate tissue growth

- Extend the characterization of soft sensor-actuator technology in a robotic implant for tissue lengthening, by establishing a set of benchmarking tests across different sensor-actuators
- Investigate further adaptive soft sensing and localisation with minimal resources, to enable information rich sensors to be integrated into soft robotic implants

## 1.4 Preview of Contributions

As a result of the work conducted towards the aims and objectives presented earlier, this work presents the following contributions:

1. Introduction of a soft sensor-actuator, a versatile building block for soft robots, and its extensive characterisation. The underlying ballooning membrane design allows for large expansion from a compact structure, as well as already being widely used in soft robotic structures. (Chapter 3)
2. Testing of soft ionic solution sensing actuators in a set of clinically relevant benchmark tests to validate the potential improved safety in robotic implants. (Chapter 3 and 4)
3. Design, fabrication and testing of multiple soft sensors, and their validation in *ex vivo* tissue experiments and wearable applications. (Chapter 3, 4 and 5)
4. Introduction of the concept of tunable, and replenishable sensors. With the easy to use and tunable conductive properties of the ionic solution filling offering application-driven customisation and allowing reusability of the sensor shell. (Chapter 5)
5. Modelling, and characterisation of a localisation methodology for fluidic sensors with minimal hardware, thus exploring more active sensors to achieve richer information using dynamics. (Chapter 5)

## 1.5 Publications

The work in this thesis includes a number of original contributions to the robotics field. The work presented has additionally led to the following publications in academic journals and conferences, listed below:

1. Jones, J., Gillett, Z., Perez-Guagnelli, E., and Damian, D. D. (2020). Resistance tuning of soft strain sensor based on saline concentration and volume changes. In *Annual Conference Towards Autonomous Robotic Systems*, pages 49–52. Springer (first author)

2. Jones, J. and Damian, D. D. (2022). A soft fluidic sensor-actuator for active sensing of force and displacement in biomedical applications. In *2022 IEEE International Conference on Intelligent Robots and Systems*, pages 6913–6919. IEEE (first author)
3. Jones, J., Pontin, M., and Damian, D. D. (2024). Pulsating fluidic sensor for sensing of location, pressure and contact area. In *2024 International Conference on Robotics and Automation (ICRA 2024)*. IEEE (first author)
4. Jones, J. and Damian, D. D. (2024). A soft fluidic sensor-actuator for active sensing of force and displacement applied to tissue probes and implants. In *Transactions on Medical Robotics and Bionics*. IEEE (first author) (**submitted**)
5. Herzig, N., Jones, J., Perez Guagnelli, E., and Damian, D. D. (2021). Model and validation of a highly extensible and tough actuator based on a ballooning membrane. In *2021 International Conference on Robotics and Automation (ICRA 2021)*. IEEE (second author)
6. Perez-Guagnelli, E., Jones, J., and Damian, D. D. (2021). Hyperelastic membrane actuators: Analysis of toroidal and helical multifunctional configurations. *Cyborg and Bionic Systems* (second author)

Publication 1 was presented orally as a short paper at the virtual conference Toward Autonomous Robotic Systems 2020, hosted by the University of Nottingham, United Kingdom. Publication 2 was presented virtually as an oral presentation at the International Conference on Intelligent Robots and Systems 2022, held in Kyoto, Japan. Publication 3 was presented at the International Conference on Robotics and Automation 2024, in Yokohama, Japan. Publication 4, is an extension of publication 2 and was submitted as an original article to Transactions on Medical Robotics and Bionics. The contributions to publications 5 and 6 are minor and presented in appendices.

The contents in Publication 2 and its further extension in Publication 4, have contributed to the content of Chapter 3. The work conducted as part of Publications 5 and 6 complement the work that makes up Chapter 4. The contents in Publication 1 and Publication 3 form the content of Chapter 5. Finally, the introductions from all the publications have contributed to the contents of Chapters 1 and 2.

In addition to the main body of work, the following publication features contributions primarily developed prior to the PhD and is attached as an appendix for reference (Appendix D):

1. Perez-Guagnelli, E., Jones, J., Tokel, A. H., Herzig, N., Jones, B., Miyashita, S., and Damian, D. D. (2020). Characterization, simulation and control of a soft helical

pneumatic implantable robot for tissue regeneration. *IEEE Transactions on Medical Robotics and Bionics*, 2(1):94–103 (co-first author)

## 1.6 Thesis Outline

The thesis structure is summarised in Fig. 1.2, and is as follows:

- Chapter 1 provides context and introduces the contributions of this work.
- Chapter 2 presents the background of the project and related work. Section 2.2 explores the developments in medical robotics and presents some unique challenges and opportunities in this field. The field of soft robotics is presented in section 2.3, and then further review of the actuators, control and sensing of soft robots are discussed in sections 2.4, 2.5 and 2.6 respectively. Section 2.7 highlights the research questions driving this work, before a conclusion of the literature review is presented in section 2.8.
- Chapter 3 is the first technical chapter and introduces a soft sensor-actuator, capable of large extension from a compact initial form, as well as measuring both displacement and force. The fabrication, modelling and the results are presented in sections 3.3, 3.4 and 3.5 respectively, before the investigation into two biomedical applications that are presented in section 3.6.
- 
- Chapter 5 introduces a soft strain sensor with tuning capabilities and a soft pulsating fluidic sensor with localisation capabilities. The fabrication procedure, the modelling and the results of the tunable strain sensor are expanded on in sections 5.3, 5.4 and 5.5 respectively. For the soft pulsating fluidic sensor, these same parts are explained in sections 5.8, 5.10 and 5.11 respectively, with additional applications demonstrated in section 5.12.
- Chapter 6 concludes the thesis, detailing the future work needed to address the limitations of this work, as well as providing an outlook on the future of the field.

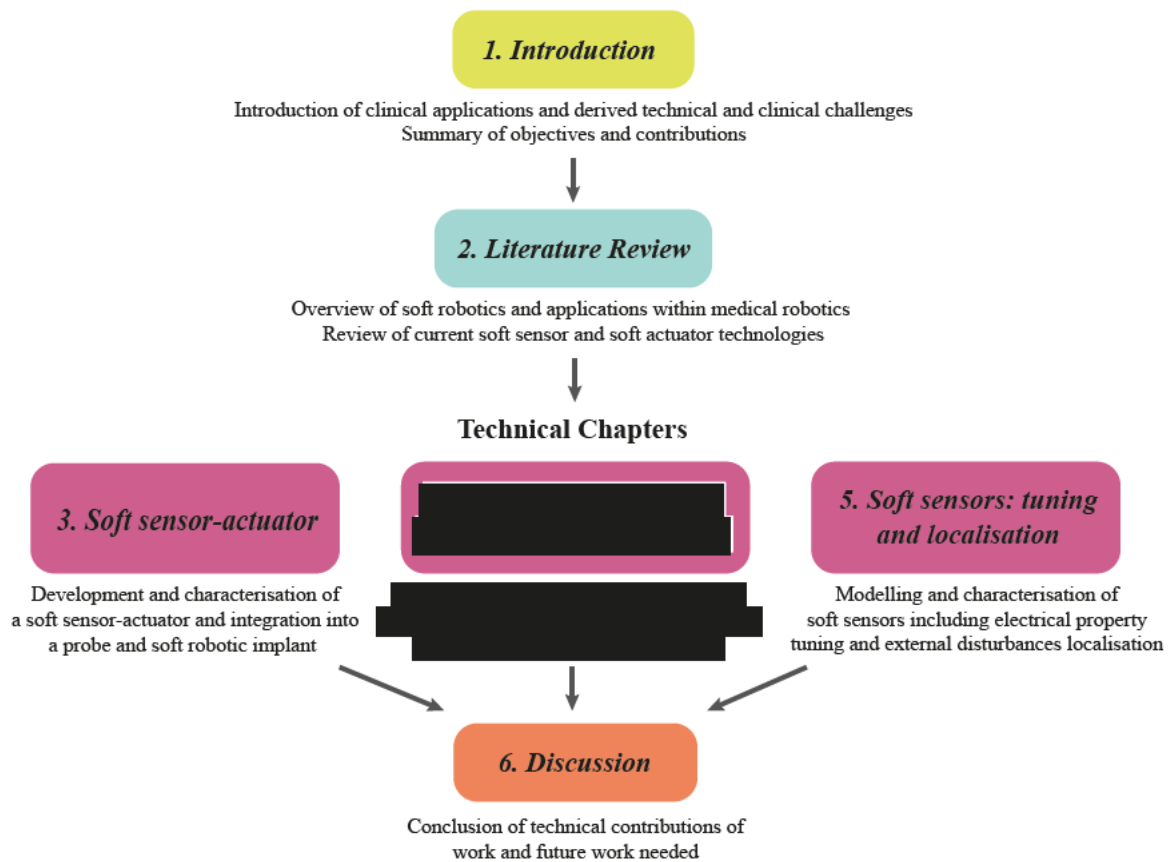


Fig. 1.2 Thesis outline



# **Chapter 2**

## **Literature Review**

### **2.1 Preface**

Making a safe soft robotic implant requires an understanding of the intersecting medical and soft robotics fields, overcoming the stringent requirements for medical devices and leveraging the potential of new robotics technologies to pave the way towards improved patient care. With this in mind, this literature review will aim to give a complete and concise overview of these two fields. The review will begin by looking at the medical field, including the robotic technology already being used and the current research efforts being made. The need for more compliant mechanisms will introduce the field of soft robotics, where the major breakthroughs in soft robotics will be explored. From there, some common soft actuation techniques will be shown, and their drawbacks explored. The focus will then move to the control and modelling aspects of these robots, which are complicated by their infinite degrees of freedom. Finally, the soft sensing solutions currently being employed will be presented, before highlighting the research gap for this work.

### **2.2 Medical Robotics**

Robots have started to be involved in many aspects of daily living, becoming more ubiquitous in the workspace and our homes. In the field of medical technology, they are helping to improve patient care both in and outside hospitals. Within the field of medical robotics, there has also been a move towards using technology to carry out medical procedures in the body, and giving them more autonomy in safety critical cases. In this section, some breakthrough robots will be presented, alongside some exciting research projects being

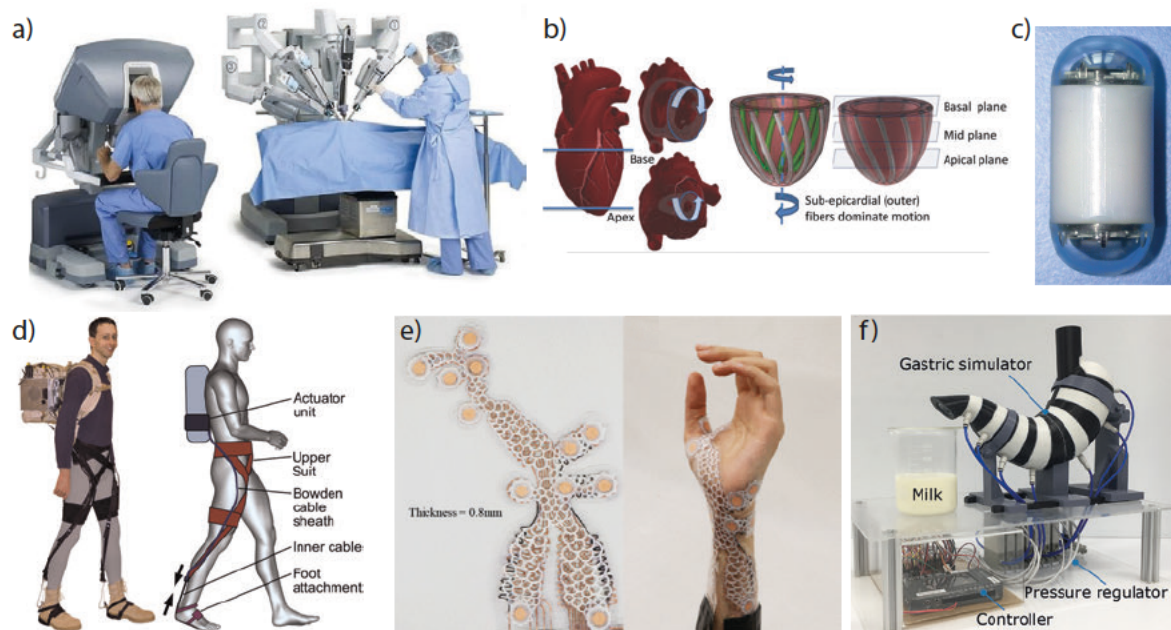


Fig. 2.1 Medical robotics. (a) da Vinci, the most widely used surgical laparoscopic robotic system (Used under the Creative Common License). (b) Example of a bioinspired heart actuator [154] Copyright, 2014, John Wiley and Sons. (c) Pillcam, a swallowable diagnostic tool for the gastrointestinal tract [52] Copyright, 2004, John Wiley and Sons. (d) Exosuit designed for walking assistance. Reprinted from [3] with permission from SAGE publications. (e) Soft pressure skin to detect pressure and supplement hand orthosis. Reprinted from [177], ©2020 IEEE. (f) SoGut, example of a gastric simulator [43], reprinted with permission. The publisher for this copyrighted material is Mary Ann Liebert, Inc. publishers.

undertaken. The focus will mainly be on surgical robots and robots that help replace or enhance body functions, such as prosthetics, implants and wearables.

## Wearable Robots

In the field of prosthetics and orthotics, robotic technology has found many uses in adding functionality and making the device more useful and human-like. For the lower body, the development of exosuits has transitioned from rigid designs to softer designs allowing more natural movement for the user. For example, Walsh et al. [130] have been working on soft exosuits to help with load carrying tasks, as well as adapting this technology for stroke patients, to assist with their recovery [6]. Both case studies are promising, although for the stroke patients, long-term impacts still need to be studied. Besides augmenting functionality, work on replacing lost limbs in the form of prosthetic legs [5] has also been investigated, working towards natural walking without the metabolic burden usually incurred.



There has also been work on assistive soft robots for the upper body, including soft haptic devices [1], to provide force and direction cues to the skin on the arm. There has been work to reduce discomfort and pressure sores of patients wearing orthoses, by monitoring high pressure point to check for a good fit [177]. Additionally, softer prosthetic arms are starting to be researched, in the hope that they will give users more dexterous manipulation [209]. Beyond daily use, as part of replacing/enhancing limbs, robots are starting to have a more prominent role in the operating room, as well as tool to help with diagnosis.

### **Artificial Organs and Simulators**

The development of artificial organs and simulators is paving the way for the next generation of surgeons and engineers designing medical devices. From an engineering perspective, these simulators are providing more realistic environments for engineers and designers to test their medical devices in. Stomach and intestinal simulators are allowing engineers to test new capsule robots and drug delivery robots, as well as improving their fundamental understanding of the human body [67][12]. In the field of tissue engineering, a variety of bioreactors [170][121] and tissue simulators [141] are being used, and created, to test new stimulation protocols and more realistic strain conditions. The accumulation of progress across these fields is kickstarting the production of fully artificial organs, to address the current shortage of organs for transplant.

From a surgical perspective, artificial organs and simulators are aiding in moving away from animal studies and providing more realistic and accessible training environments [97]. Simulators give both experienced and trainee surgeons better surgical practice, by allowing them to run-through surgical procedures using patient-specific information. For junior doctors, these simulator tools, also allow for more standardised learning, as well as greater diversity in the patient cases observed, and objective performance metrics during their training periods. These developments are also expanding beyond medicine and into veterinary sciences to transform the future for veterinarians too [8][36].

### **Minimally Invasive Surgery**

With the approval of the da Vinci surgical system at the turn of this century [175], robots for minimally invasive surgery have slowly become more mainstream. With shorter recovery times, they have become the go-to instrument for some surgical interventions. However, the technology requires long training periods for the doctors, so many groups have focused on enhancing the technology, and producing cheaper and more intuitive alternatives [185]. There has also been the development of additional tools, including more flexible ones to

allow for better visualisation and manipulation. In line with reducing invasiveness, there has also been a move towards using natural orifices to pass instruments through.

Flexible endoscopy is one such minimally invasive technique, using natural orifices to carry out imaging or therapeutic procedures. The field has seen the development of more flexible tools, designed to overcome sharp bends in organs, and thereby minimise damage considerably. There has also been the development of the PillCam [52], which provides a swallowable alternative to imaging and diagnostic endoscopy with no sedation needed, although compromising the control of the surgeon for the visualisation. In addition, research is now also focused on creating robots to crawl through the body to target locations [74], giving surgeons more control over their field of visualisation and allowing biopsies and simple procedures to be undertaken remotely [115].

## **Implantable Technology**

In both robotic-assisted and endoscopic surgeries, the surgeon is still very much involved in carrying out the procedures, however there has been an increased focus on autonomy, in part to allow more long-term interventions to happen. The increased development of technology and long-term patient needs have led to robotic implants used to replace and/or enhance functions of organs inside the body. Within this research topic, there have been exciting developments into a soft robotic heart sleeve, capable of replicating the complex motion of the human heart [153]. There have equally been many developments in creating devices that could locally distribute therapeutic cells [23], as well as actuating to reduce fibrotic tissue capsule around foreign bodies [47], all using soft materials.

While moving towards more autonomous robots [201] has shown great promise, especially in surgery, with noticeable improvements in the consistency of procedures and patient safety [163], there has also been a move towards using softer and more compliant technologies. In addition to being able to conform to a given space, soft materials have been seen to induce a reduced foreign body response in contrast to their rigid and less mechanically compliant counterparts [120]. However, numerous challenges still exist in bringing the latest technological advances to the medical field.

## **Challenges**

### **Research Translation**

One of the major challenges in medical robotics is overcoming the various regulatory and socio-technological barriers to transfer research knowledge out of academia into developed

and clinical viable products. The regulatory hurdles include extensive benchmarking of test results to ensure reproducibility and patient safety [38]. This process can take years to go through the necessary clinical trials before commercialisation. In addition, there are also various socio-technological considerations, involving gaining patients' and doctors' trust to use the technology [35], from wariness of robots to poor integration into the clinical workflow adding additional burdens to medical staff. Finally, research is not always conducted with technology transfer in mind, leading to a mismatch between research and the realities of medical device development, resulting in a substantial lag between development and commercialisation, if it even occurs [49].

### **Accessibility**

Accessibility is a key problem for medical robots. While robots could help democratise healthcare and provide greater access to expert care in remote regions, using teleoperation for example, it could also further cement the global healthcare divide. Ensuring that the medical robots are affordable and accessible is a difficulty yet to be overcome, as the cost and technical expertise needed for operation is still prohibitively high. Additionally, as robotic-assisted surgery has a much greater environmental impact [196] and the use of single-use robotic components increases, the sustainability of medical robots needs further attention.

### **Robot-Tissue Interface**

Finally, from a technical standpoint, the complexity of the heterogeneous human body and finding appropriate ways to interface with the tissue is difficult. Having a good fit, to reduce discomfort, and inflammation, to patients due to friction and foreign body response [120], is an important challenge when transitioning from the heterogeneous human body to the traditionally rigid robots, such as prosthesis or implants. Poor fit can also worsen functioning, for example in exoskeletons where interfaces are known energy sinks [102], hindering the full potential of the device. Interfacing with biological tissue also imposes restrictions on the materials used, with biocompatibility, in addition to sterilisation, and general hygiene being key. Flexible and soft materials provide some answers given their intrinsic safety, and ability to adapt to the surrounding tissues, and will be explored in the following sections.

## **2.3 Soft Robotics**

Compliance can be achieved in three main ways: computationally - by implementing software-based impedance control [172]; mechanically - by using compliant mechanisms, such as

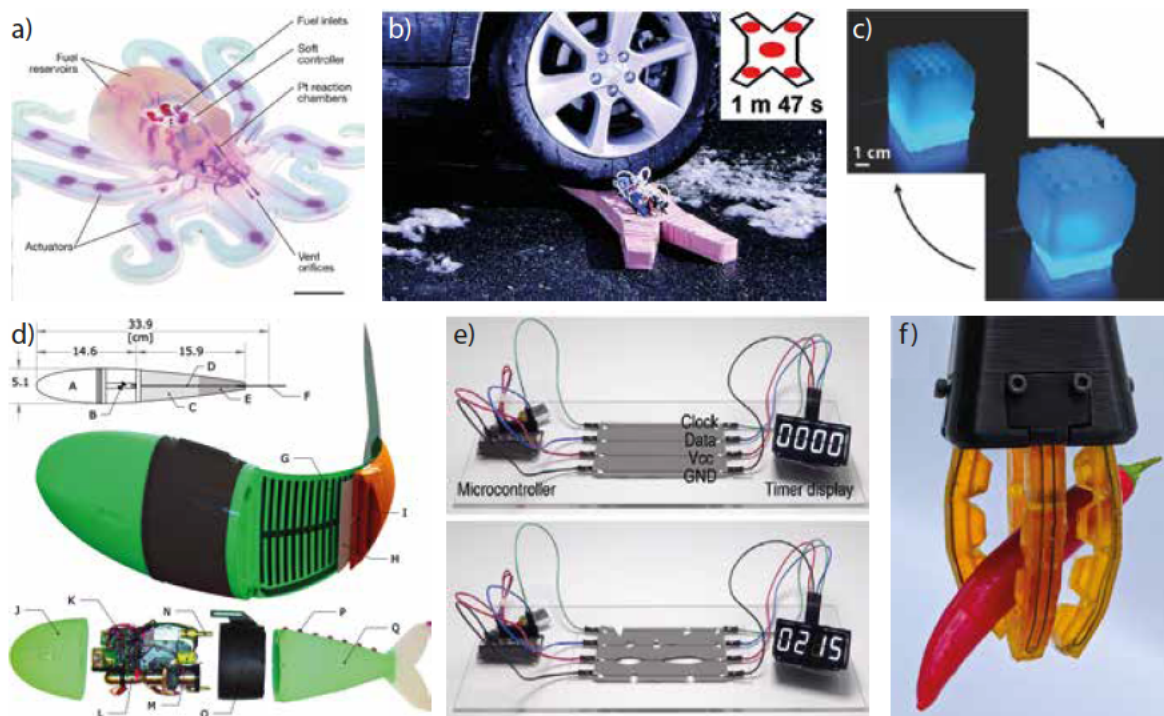


Fig. 2.2 (a) Soft octobot, one of the first examples of a fully soft robot [194], reproduced with permission from SNCSC. (b) Demonstration of inherent resilience of soft robots [183] reprinted with permission. The publisher for this copyrighted material is Mary Ann Liebert, Inc. publishers. (c) Click-e-bricks example of modular soft robots [116] Copyright, 2014, John Wiley and Sons. (d) Bioinspired soft robotic fish design [109], reprinted with permission. The publisher for this copyrighted material is Mary Ann Liebert, Inc. publishers. (e) A self-healing elastomer composite [110], reproduced with permission from SNCSC. (f) Self-healing soft robotic gripper [155] (Used under Creative Commons License).

series elastic actuators [143]; and via materials - by working with soft materials. Compliant robots, and in particular soft robots, provide many benefits including: adaptability, inherent safety, potential for complex and multifunctional behaviours, as well as being low cost and lightweight. By leveraging their elastically deformable bodies, soft robots are highly adaptive, and capable of navigating and surviving in highly complex environments [183][68]. Additionally, the predominant use of silicones in soft robotics, make them biocompatible from both a toxicology perspective, and the strong match-up with tissue properties, thereby reducing the potential fibrotic response. Finally, their soft bodies, and their resulting theoretical infinite degrees of freedom, with the body's ability to move in any direction at any time, open up the possibilities for more complex behaviours, than would be possible with more traditional robots.

Although soft robots offer many possibilities, they do present a number of drawbacks including: reduced precision and accuracy; reduced strength and durability; the need for complex programming and support equipment; as well as the current lack of adapted electronics for the field. Although the theoretical infinite degrees of freedom of soft robots allow for complex behaviour, they do increase the complexity of modelling and programming needed to provide control, so the end behaviour is often less precise and accurate. The soft materials used are equally prone to wear and tear, reducing the robot's lifetime. Once these weaknesses are addressed, the full potential of soft robotics can be fulfilled and will provide new tools in a number of different fields.

The focus for the remainder of this section will therefore be on these soft robots, including the research areas of soft robots, their fabrication, and integration. Then going into an exploration of the actuators, their modelling and control, and the sensors used in the following sections.

## Research Areas

Soft robotics is at the intersection of many fields and lots of different research directions have been explored: biological and medical devices; micro-robots; and origami fields, where material selection is already widely considered; as well as material science, where 3D printing of soft structures are being explored. Throughout, there has been a large influence from bioinspiration, driven by the desire to replicate natural soft systems such as the octopus tentacle, the caterpillar crawling motion, etc.

These developments are also happening at many scales, including: large, with growing soft robots that are able to navigate environments [68]; small, with robots capable of multi-modal locomotion by exploiting the extra degrees of freedom and compliance to increase the modes of locomotion [76]; and very small, including self-assembling SoftCubes able to reconfigure into multi-layer structures [203].

The potential for soft robotics equally spans many applications. In the medical field, including examples given above, soft robots provide an ideal platform for interacting more efficiently and safely with the human body. In industry, human-interacting robots have a need to be soft to not endanger workers on the factory floor, as well as grippers needing to be soft and have more universality for the manipulation of a greater variety, including softer and more delicate, objects on the production line [17]. Consumer electronics are also seeing uses, with the development of soft wearable technology. Finally, in the field of biomimicry [109], research into soft robots is helping to understand invertebrates and more widely the interaction between soft tissue and rigid bones of all animals.

Each of these applications also impose specific requirements on the design. Within the medical field, using biocompatible materials and meeting the anatomical dimensional restrictions is vital. Controllability and tunability are key features to ensure safe and controlled responses to external stimuli, and a natural morphological adaptation to changes of the host environment. Finally, portability of the system is essential to ensure ease of use outside of hospital, and in home settings. All of these pave the way towards more personalised medicine, and patient-specific robotic implant solutions, by combining the inherently safe and compliant nature of soft materials with novel tissue engineering approaches. However, challenges pertaining to soft robots, and in particular soft medical robots, still remain.

## **Challenges**

### **Fabrication**

One of the key challenges for soft robotics lies in their fabrication and manufacturing. Consistency and reproducibility are the shortfalls of soft robotics and are the key barriers towards their wider industrial adoption. The rapid growth of 3D printing has laid the foundation for more scalable and repeatable manufacturing of soft robots, as well as allowing the seamless combination of materials, however the field still heavily relies on lamination and lithography techniques for thinner devices and for prototyping.

As mentioned above, two more common, and still widely used, techniques for soft robot fabrication are lamination-based approaches and soft lithography, which involve using masks and moulds. These techniques are fast, and easy to use, and can be employed to make devices from a few millimetres thick - such as sensors, to much larger ones [64]. However, one of the downfalls of these methods is the creation of complex shapes, which require time-consuming multistep processes. The dependence on manual work also leaves room for error, creating repeatability problems, which is where alternative fabrication techniques can come in.

With the significant developments in 3D printing, there has also been a move to 3D printing soft structures [26]. By eliminating the often multistep process, the hope is to ease manufacturing, as well as achieving better standardisation. 3D printing, also allows different soft materials to be printed at the same time, allowing more complex structures to be made [171]. However, one of the drawbacks of 3D soft printing is the difficulty in printing silicone elastomers, which are liquid before crosslinking, often resulting in more rigid and easily printable soft materials being used, thus limiting the achievable softness of 3D printed soft robots.

To increase the prototyping abilities with soft materials, there has also been some initial work towards creating easily fabricated standard blocks which can be reconfigured. One

such structure is ‘click-e-bricks’ [116], which are similar to soft Lego bricks, that can be assembled in different configurations to achieve different functionalities. There has also been work into developing a universal soft robotic sheet, Omniskins, that can be wrapped around inanimate objects to give them function [14].

### **Integration of Sensors, Actuators and Control**

With the fabrication techniques above, there have been developments towards fully soft robots, that integrate soft sensors, actuators, controllers and power sources, and are not tethered to rigid components. This is an ambitious goal and some significant work in this area includes the octobot [194], an entirely soft robot capable of generating its own power, actuation and sensing, even if its working life is short and its actuation sequence is simple. Other works to achieve more complicated soft control, includes using microvalves and multiplexing, to switch between actuation channels and actuate multiple channels simultaneous [211], but still rely on an external pressure source. In [58], the authors introduced the concept of a soft matter computer, using fluid flow and different pulses of fluid, to achieve digital and analogue computing and demonstrated its use in multiple applications. One of the major drawbacks was the speed of the computation, excluding many real-time applications, as well as not including any sensing.

To bypass the challenge of achieving control without electrical or pneumatic connections to external equipment, some groups have looked towards leveraging biology. To start, one exciting research area is going beyond printing soft structures to the inclusion of biological systems and engineered cells within the soft robots to act as sensing devices and enhance the capabilities [87, 171]. The inclusion of cells allows the robot to have functions that might not have been possible before in a more compact form, however keeping the cells alive and integrating them well is extremely challenging.

In addition to the inclusion of cells, there has been the development in other aspects of robotics, such as the power source and the computation, helping bridge the gap towards entirely soft, autonomous and untethered robots, rather than just soft robotic elements still relying on rigid components. The use of low-boiling point fluids, with pressure levels that can be tailored by the volume of loaded liquid, have been presented in [59] as soft power sources. There has also been work using a dielectric chamber as a source for a hydraulically driven soft actuator [205]. Furthermore, work integrating sensing and actuation into a single channel in soft hydraulic actuators has been conducted [71].

## Resilience to Faults and Damage

Resilience in robotics is not a new concept with many efforts already relying on redundant components and modular architectures [207]. The field spans from relying on: algorithmic resilience - through robots that change their behaviour; to architectural resilience - with reconfigurable robots; and via material resilience - using soft robotics. Soft robots in particular are naturally resistant, withstanding mechanical stresses and extreme cold and hot conditions. [183]. However, they are still prone to different kinds of faults: material degradation; fabrication failures; overpressurisation and bursting from sharp objects. Therefore, the need to overcome these faults is crucial, with several approaches taken to overcome the limitations of soft robots ranging from compensation control strategies, additional actuators and self-healing materials.

In terms of algorithmic resilience, some efforts have been made in determining strategies for dealing with certain faults, after first identifying the risks and the consequences of errors happening [9]. There has also been work in detecting damage using grid sensors in combination with pressure sensors, and adjusting pump inputs to compensate for puncture damage [111], or isolating channels using drops in pressure during bursts [142]. Machine learning has equally shown potential to identify and overcome faults [27], as well as the more exploratory work, taking inspiration from neuroscience approaches, to mimic the concept of pain [133]. However, for extensive damage, these compensatory behaviours are not always sufficient, and are equally more geared towards electrical faults.

Using the structure of the robot is another method to increase resilience. Some robots make use of additional actuators, to help replace functionality if one is lost [9]. This method is notably used in octopod and hexapod robots (which adjust their gait) and is poorly explored for their softer counterparts. Another approach is reconfigurable and swarm robots, which make use of redundant units and can adapt their collective behaviour in case of failure [203]. This approach is suitable for systems with enough space and/or where several units can be released.

In contrast, self-healing approaches aim to fix the problem, rather than circumvent it [13]. Work in this direction includes the development of self-healing soft pneumatic robots, made of a fully recyclable self-healing polymer [179], and the use of hydrogels as naturally self-healing materials [103]. However, this work, necessitates the robot deflating and an additional heating source to heal. Additionally, the robot recovers functionality, but not always aesthetically. Another approach in this field is the use of conductive liquid for reconfigurable electronic circuitry [110]. Using this technology, when the robot is punctured, the electrical pathways are able to reconfigure and the oxidising layer of the liquid ensures that no leaking occurs. The same research group also successfully tried it in a soft quadruped that



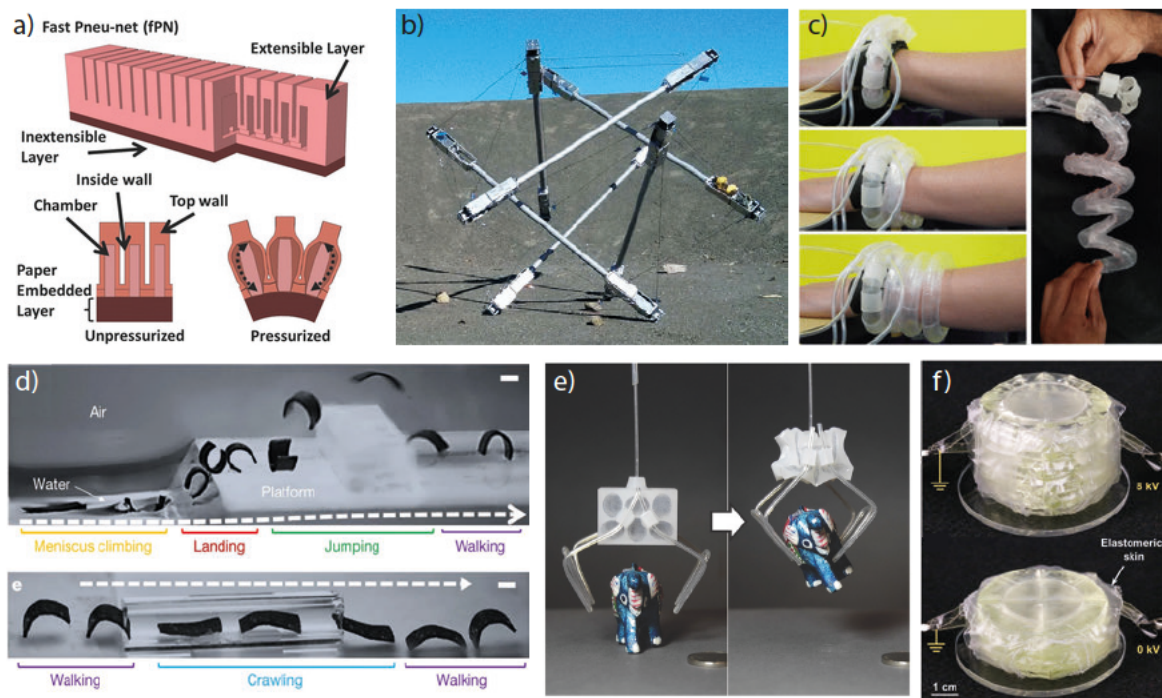


Fig. 2.3 (a) Pneumnet actuators, one of the more common soft pneumatic actuators [117] Copyright, 2014, John Wiley and Sons. (b) Tensegrity robot for space exploration. Reprinted from [161], ©2015 IEEE. (c) Haptic wearable, that can conform to users forearm. Reprinted from [1], ©2018 IEEE. (d) Small-scale soft-bodied magnetic robots [76] reproduced with permission from SNCSC. (e) Example of an actuator exploiting auxetic materials [200] Copyright, 2015, John Wiley and Sons. (f) A HASEL actuator using electrostatic zipping to achieve fast and large deformation actuators [114] (Used under the Creative Commons License).

was exposed to hole-punch damage without compromising its crawling behaviour. However, the processes and materials used for self-healing are not always biologically friendly.

## 2.4 Soft Actuators

Fundamental research into soft actuators has been made with the aim to make strong, lightweight and cheap actuators. Lots of developments have been made using a variety of different technologies [16], which will be looked into in the following subsections.

## **Actuation Technologies**

### **Flexible Fluidic Actuators**

Some of the most popular soft actuators are fluid elastomeric actuators, including the widely used Pneunet actuators [117]. In these actuators, the flow of liquids or gases (hydraulics and pneumatics) is used to drive changes in the shape of a robot, to be used for manipulation [77], locomotion [164], etc. They have the advantage that the electronics can be placed far from the site of actuation, although they are prone to bursting, and can exhibit highly nonlinear behaviour, due to the compressibility of gases and the dependence on temperature.

### **Tendon-Driven Actuators**

Tendon-driven robots are equally popular, due to their high strength. However, this strength comes at the cost of forfeiting softness by using rigid components to tighten the tendons. Nevertheless, they have shown great promise in applications such as exoskeletons, where the tendon provides enough compliance to allow natural walking, whilst still having enough strength [130]. Tendon-driven actuation has also been used for the development of SUPER-ball, an untethered tensegrity robot [161]. This robot was designed with NASA for use in space exploration, where tendons provide an optimal solution for compact and strong locomotion and enough compliance to move over highly unpredictable environments.

### **Dielectric Actuators**

Dielectric actuation is also common and is achieved by charging electrodes separated by an insulating layer. When charged, the electrodes are attracted to each other and squeeze the dielectric material in between. This method is very quick, although it necessitates high voltages, which are not always suitable. Nonetheless, they have been shown to be simple and light actuators, and when combined in lattices with others, similar to muscle structures, they can be very powerful [176].

### **Magnetic Actuators**

Magnetic actuation has been investigated, using magnetic fields to generate motion in soft materials. This method is very fast, can be fully actuated in less than 1 second, and can achieve complex shapes, including 2D and 3D structures [96]. The other advantage is that the driving mechanism can be completely separated. This actuation technique has also been used for polymers with magnetic particles to achieve different modes of locomotion [76], but

the force generated can be quite low, and therefore this actuation is potentially more suitable for dexterous tasks.

### **Heat-Driven Actuators**

Apart from the more well-established soft actuator types, a variety of more novel actuation methods, driven by heat, are being explored for soft robots. One such type are heat-responsive actuators, such as shape-memory materials, that rely on heating to switch between a rest and active state. These often drive complex shape changes, but are not always strong, and the resting state configuration can be unpredictable. Beyond these heat-responsive actuators, developments towards photo-thermal actuators, taking inspiration from plants [191], can be seen, but often come at the price of being much slower. Finally, growing research into hydrothermal actuation, exploiting the water capture and evaporation potential of hydrogels, is also being translated into soft robotics [198].

### **Other Actuation Technologies**

To finish, a plethora of unconventional soft robotic actuators are also being investigated, ranging from acoustically powered soft robots [89], to robots leveraging on the neighbouring biological, and chemical, fields to achieve actuation. Acoustic actuators have the potential to be the basis for all-mechanical soft robotic systems, but they are still in the early stages of development, and a lot more work is needed to achieve more complex actuation. Building on the chemical field, chemical reactions, including the explosive combustion of butane [107], have been demonstrated as viable sources of actuation, with high-density power generation. However, their resulting movement can be unpredictable, and they have a limited lifespan. Similarly to combustion, actuators exploiting mechanical instabilities, and stored energy in materials, have been explored [129][61], but again tend to be unpredictable. Harnessing biology too, can produce highly complex actuation, but with a high cost of complicated fabrication.

## **Challenges**

### **Complex and Multifunctional Behaviours**

On top of using a single technology, some research groups have also investigated combining technologies to harness the advantages of each, and attempt to overcome their limitations. One example is the Peano-HASEL actuator, which makes use of the fast actuation of dielectrics and the strength of hydraulics to produce lightweight and cheap actuators that

have good strength [90]. There has also been work on the inclusion of liquid metal in liquid crystal elastomers [56]. This work showed success in combining the reversible behaviours of liquid crystal elastomers, and the good thermal and electrical conductivity of liquid metal, to achieve complex and multifunctional behaviour.

### **Strength and Durability**

The intrinsic softness of soft actuators often comes with the drawbacks of less strength, and worse durability. To overcome these limitations, groups have explored a variety of different methods to help reinforce targeted areas, by integrating tactically placed fibres [34] or origami shells [127], increasing the overall structural strength of the actuators. Besides the benefit of strength, these reinforcements also help control the actuation, by limiting the degrees of freedom, however, the reinforcements can lead to weaknesses in other areas of the actuator. While, mainly explored with pneumatic actuators, some groups have also extended this idea to hydrogel actuators, where fabric wrapped around the hydrogels helped both their load-bearing capabilities and avoided them becoming too brittle [122].

### **Portability**

Finally, one of the considerations and limitations of current soft actuators is their portability, with many tethered to power sources, therefore limiting the ability of soft robots to explore remote locations [86]. One pioneering research work, bridging the gap for pneumatically-driven soft robots, and moving away from integrating electronics in the soft body itself [184], is an untethered legged robot [48], using a ring-oscillator pneumatic circuit and a constant pressure source from a  $CO_2$  canister to power four legs. Although still relying on a rigid canister, it remains a significant move away from the electronic pneumatic circuits traditionally used. One actuator type already considered to be untethered, are magnetic actuators. Although not physically tethered, they require significant external infrastructure to power them, and therefore do not truly present an untethered and easily portable actuating solution. The chemical reactions presented above or the use of actuators driven by environmental conditions, such as humidity or light, are increasingly being studied, but these are then tied to the environment or the limited supply of the reactants.

## **2.5 Soft Robotics Control and Modelling**

In terms of modelling and control, several approaches have been taken [159]. The control approaches can be split into three main categories: embodied, model-based and data-driven.

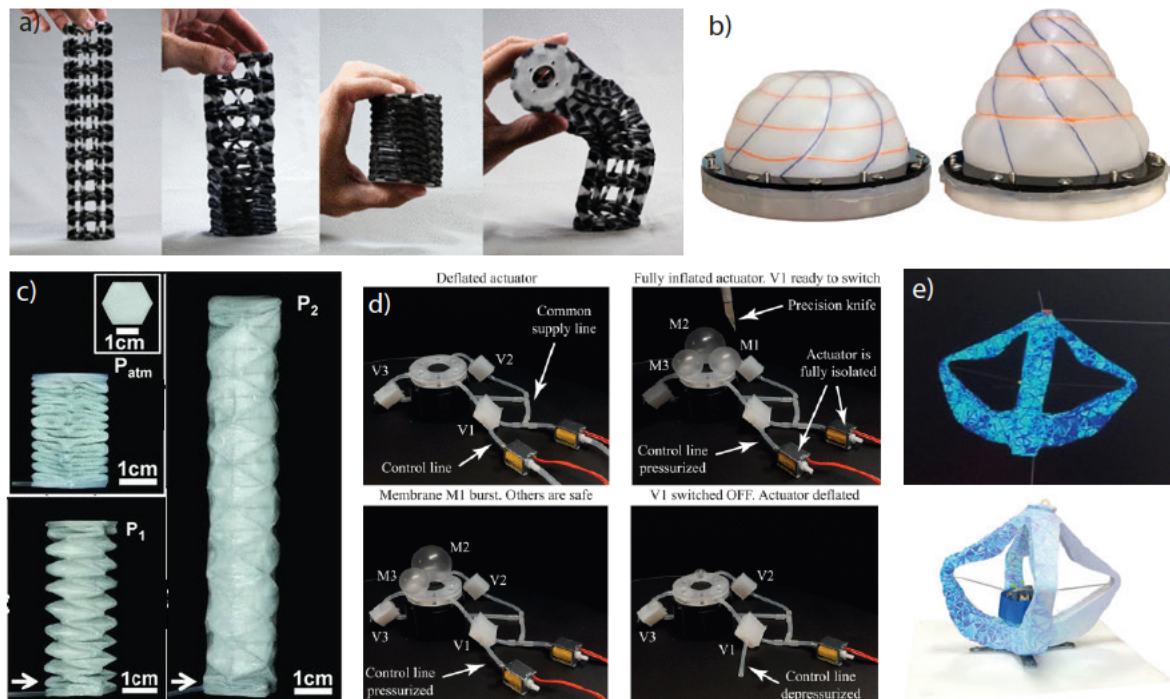


Fig. 2.4 (a) Multi-material printed origami tower. Reprinted with permission from [106] © 2017 IEEE. (b) Example of constraints designed to control inflation of elastomeric membrane. Reprinted with permission [167]. (c) Soft actuator with embedded paper origami shell to constrain lengthening [112]. Copyright, 2012; John Wiley and Sons. (d) Demonstration of fault-tolerance embedded into design of fluidic components. Reprinted with permission from [142] © 2022 IEEE. (e) Example simulation using SOFA, software toolkit for soft robots. Reprinted from [33] with permission from Taylor & Francis.

The research focus varies from adapting traditional control strategies, to deriving controllers from data, to simply changing the morphology to constrain the robot behaviour.

## Control Approaches

### Morphological Computation/Embodied Intelligence

One more recent way of looking at passive control in soft robotics is trying to exploit the soft body to carry out some control tasks. This is inherent to the design of soft robots aimed for specific tasks and can be achieved through constraints and morphology. One example is the development of origami shell reinforcements to control the bending and improve reliability [127]. Another example includes varying the placement of fibre reinforcements to change the behaviour of a soft pneumatic actuator [34]. With variations in the number of reinforcements, as well as the angle and orientation, the group achieved extension, expansion, twist and

bending and demonstrated the segmented use of different patterns to replicate the motion of a human finger. Work has also been carried out exploiting pressure differences during bursts of pneumatic actuators to embed fault-tolerance into the design of pneumatic valves [142].

### **Model-Based Control**

Besides outsourcing control to the body, work has been carried out to develop models to understand the behaviour of soft robots better. This improved understanding could allow the crossover to applying well-established rigid body control methods to soft robots [11]. Over the past century, there have been many groups focusing on understanding the material properties of hyperelastic materials. Lots of work is still being carried out, in particular looking at the pre-stretching of the membrane [108], with most work concentrating on looking at the inflated membranes with no external contacts. Meanwhile, greater understanding of the nonlinear pressure-volume relation has enabled research groups to exploit this to trigger fast snap-through ballooning responses, using combinations of different balloons and constraints [126]. However, the highly nonlinear nature of soft robotics drives the need for more continuous and dynamic modelling. To this end, some groups have conducted work on creating a unified framework allowing for modelling, simulation and control of soft robots [33]. By combining finite element modelling (FEM), traditionally reserved for modelling given its computational costs, with optimised inverse models based on the derived equations and applying traditional control algorithms, the research group has shown the potential to achieve real-time closed-loop control of soft robots [21].

### **Model-Free Control**

In contrast to methods aiming to understand the fundamentals of the behaviour, data-driven approaches have become more popular with the developments in machine learning. By using large amounts of data, the hope is to extract behaviour patterns. In [27], a summary of the works carried out tackling the problem at different levels is presented: learning the inverse kinematics - limited by a quasistatic assumption; learning forward dynamics - enabling the use of predictive control; or direct learning of controllers - although this requires a large amount of data, and time, compared to the other two methods.

In terms of machine learning approaches, one key work explored perception using neural networks and soft sensors [181]. Taking inspiration from the sensory architecture in the human body with a large amount of redundancy, they presented a time-dependent model using machine learning. Their method can be used for different sensors and considered the drift and hysteresis characteristic of soft sensors. However, the main drawback of machine

learning approaches is that they give no physical intuition and the controller is tied to a given data set, not often translating well to other robots. One approach, that blends model-based and model-free approaches is nonlinear system identification, identifying parameters of a model. This approach gives some intuition of the system and reduces the training time, but requires a good initial model, often makes assumptions, and constrains the behaviour [79, 18].

## Challenges

### Infinite Degrees of Freedom and Hyperelasticity

An inherent challenge of soft actuators is controlling the infinite degrees of freedom resulting from the hyperelasticity of the materials used. The soft robots natural adaptation to the environment, one of the key strengths of soft robotics, also exacerbates the uncontrollability with the body of the soft robot interacting with the environment in unpredictable ways. In model-free control, where the controller is derived from training sets, the controller, when exposed to new scenarios, can exhibit uncontrolled and unpredictable robot behaviour. Model-based controllers also struggle in these scenarios, while not tied to training data, they still rely on certain modelling assumptions and current dynamic theory understanding with its limitations in faithfully modelling hyperelastic materials. Therefore, in complex, real-world applications, the embodied intelligence approach highlights the opportunity of benefitting from good designs to make control tasks easier and constrain the possible behaviours of soft actuators.

## 2.6 Soft Sensing

Soft sensors ideally give information on the state of the system, including itself and its surroundings, without compromising its movement [181]. The need to provide information, without hindering robot motion, is the underlying principle in the field and led to Wang et al. [190] deriving the five requirements of soft sensors: compliance; limited dimensionality and spatial distribution; resilience and durability; robustness to environmental interactions; and easy integration into the robot, by using similar fabrication methods and materials. The internal and external inputs to be monitored include bending, twisting, strain, and detection of objects, as well as combinations of these different disturbances. Some popular technologies used are detailed below, as well as the overall challenges of soft sensors, with a particular focus on the needs for biomedical robots, where size and biocompatibility are among the top considerations.



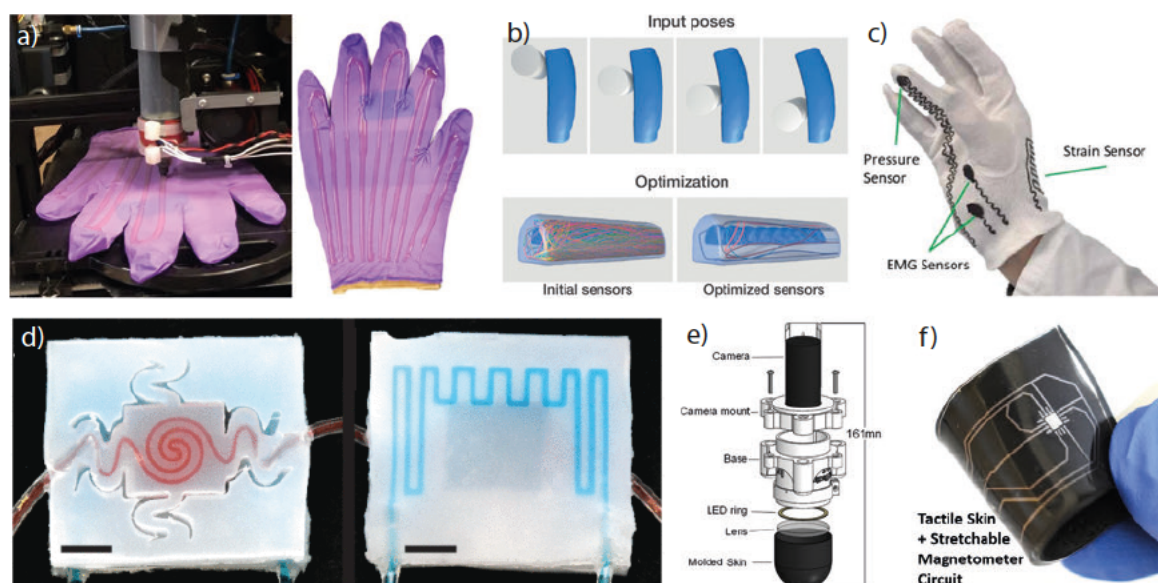


Fig. 2.5 (a) Demonstration of a printable self-healing and sensing hydrogel [66] (Used under the Creative Commons License). (b) Example of targeted sensor placement [178], reprinted with permission. The publisher for this copyrighted material is Mary Ann Liebert, Inc. publishers. (c) Example of a 3D printed graphite based sensor for hand monitoring [134] (Used under the Creative Commons License). (d) Sensor capable of simultaneous pressure and strain sensing using strategic channel design. Reprinted from [62] © 2020 IEEE. (e) TacTip, widely used soft optical sensor [192], reprinted with permission. The publisher for this copyrighted material is Mary Ann Liebert, Inc. publishers. (f) Magnetic sensor using embedded magnetometer [70]. (Used under the Creative Commons License).

## Sensing Technologies

### Resistive Sensors

One of the more popular methods is resistive sensing, which makes use of conductive materials and geometry changes (length and cross-sectional area) to detect strains. This technique usually makes use of embedded conductive liquids or textiles [206, 160, 111], with more recent work also including conductive hydrogels or other printable conductive inks [66, 134] making them easy to integrate into robots or gloves and other clothing. The sub-category of ionic resistive sensors, lends itself particularly well to medical applications, given the opportunity to make inherently biocompatible sensing mediums from saline solution based sensors. The use of fluid also allows using it as an actuator, with Helps and Rossiter demonstrating using ionic liquids to drive and sense stretching and bending flexible fluidic actuators [71]. Xu et al. also showed how to measure strain and force in a sensor using a channel filled with a conductive electrolyte solution and a force indenter [197], while Gupte et al. sensed simultaneous pressure and strain through the use of two distinctive overlaid



channels filled with an sodium ionic solution. The simplicity and low-profile of the sensing, with minimal and readily available external hardware, also makes them ideal for applying to a variety of applications, such as in cell cultures [105]. However, these sensors do struggle with hysteresis and nonlinearity, and are sensitive to temperature. These sensors are also prone to drift, and for ionic solutions evaporation through permeable soft membranes can pose an issue, making them challenging to incorporate within long-term medical robots, where reliable measurements are essential for therapeutic efficacy. Additionally, for sensing of different stimuli separate resistive channels are usually needed, and the localisation of stimuli requires additional electronics and sensing electrodes.

### **Capacitive Sensors**

Capacitive electrodes are another popular type of sensor and consist of two electrodes with an insulating layer sandwiched between them. When pressed on, the two electrodes come closer together varying the voltage across them. With clever configurations, they have been demonstrated to be capable of measuring pressure, as well as shear forces [151]. Overall, these sensors have good sensitivity and fast response time, but are sensitive to their environment and have more complicated fabrication steps. This makes them less widely used in the soft robotics field generally, beyond the tactile sensing applications where they are well established. Within the medical field, these sensors are therefore more suited to robots outside of the body, given their sensitivity to the environment, and less suited to implantable technology.

### **Inductive Sensors**

Inductive sensors exploit electromagnetic induction to detect or measure objects, and in soft sensors can be used to monitor a variety of aspects of soft robots including deformation, strain, and pressure. One key work includes "smart braids" [53], where pneumatic artificial muscles were reinforced with conductive insulating wires, allowing strain sensing during contraction. Other work includes mapping the flow of liquid metal over a coil to detect bending and normal forces [63]. Due to the complexity of electronics needed for signal conditioning, their use is not yet widespread in soft robotics. These sensors are also more commonly used for exoskeletons and prosthetics for medical robots, where there is more space for the required electronics.

### **Magnetic Sensors**

In contrast, magnetic sensors make use of a magnetic source and sensor, both encompassed within a soft body. Although they do suffer from magnetic interference, they have shown promise in detecting the location and the force being applied [69]. While compact and low cost, they do require the integration of electronics close to the area of sensing, making them less suitable for applications with high strain or if the electronics are more isolated from the soft body. Within small areas they have a good ability to reconstruct shapes of magnetic soft bodies, even over long periods of time thanks to the advanced electronics used, but for implantable technology the incorporation of rigid electronics does reduce their compliance and mechanical biocompatibility.

### **Optics-based Tactile Sensors**

Optical sensing uses variations in light to detect changes and has the advantage of being immune to electromagnetic interference and environmental contaminants. A number of different approaches have been explored. One key work is on the integration of optical waveguides into a soft prosthetic hand [209]. In this work, the group demonstrated strain sensing, for curvature and elongation; and force sensing, to detect shapes and softness. Another approach in optical sensing has been camera based. TacTip is one example [29], which is inspired by the deformation of skin layers on human fingers. The sensor detects objects by tracking pins on the inside surface of the membrane that change orientation with interaction. Another key camera-based technology is GelSight [204], which makes use of reflective paint and a gel layer to make objects pressing on the sensor visible to the camera. Overall, optical sensing suffers from requiring complex electronics. For the waveguides, they are also less stretchy making them unsuitable for large strain applications and the camera-based options require rigid cameras. All these considerations make them less suited for the compact yet compliant requirements imposed by soft implantable robots.

### **Other Sensors**

There has also been the development of a variety of other soft sensing methods. Air flow sensors have been used to detect changes in cavity to detect location and force magnitude of contact on a 2D planar surface with a good resolution [123]. Acoustic sensors have also been used for tactile sensing in a robotic finger [212]. There have also been many other sensing technologies including triboelectric and chemical sensors, not yet widely applied to medical applications, but with great potential for long-term and remote self-powered sensing ideal for implantables.

## Challenges

In regard to soft sensing, a number of challenges have been identified [190, 173], due to the nature of soft materials and applications they are used for. Some of these challenges are explored in more details in the following subsections.

### Multi-Modal Sensing

While a lot of focus has been on either understanding the robot or the environment - proprioceptive and tactile sensing - little work has been done towards the integration of both. One key challenge to overcome is the decoupling of the signals originating from the robot and the environment, made only more challenging by the infinite degrees of freedom of the robot. Making use of the undeniable link between the body of soft robots and their sensing capabilities can facilitate differentiating and amplifying sensing signals [169]. Some notable works harnessing this idea include Gupte et al. [62], who worked on simultaneous pressure and strain sensing using a multi-layer sensor. The group managed to distinguish the signals by using stacked layers with sensitivity in different areas, as well as the addition of appendages in the material to reduce the pressure sensor's sensitivity to strain. This idea of decoupling is shared by Noda et al. [125], who present two identical parallel channels, one with an added bump, to detect strain, bending and contact force, by using the bump-less channel as a reference to differentiate between signal sources of strain and force. Combining different sensing elements was explored in [95], where ionic liquid sensing, optical sensing and conductive fabric sensing were merged into a compact structure, capable of differentiating stretching, compression and bending deformations, although this increased the cost and added complexity to the sensing system.

### Sensor Placement

One key consideration across the different methods is sensor placement. Given the impossibility of having an infinite number of sensors, the need to optimally place them is crucial. In [37], the group proposed a method of first studying and modelling the motion of the robot and then placing the sensors to locations that underwent the most strain. The group achieved good results in differentiating between serpentine and twisting motions, identified as key motions for soft roboticists, although this method relies on knowing robot behaviours. Other groups have taken a more general approach [181], making use of a bioinspired vascularisation design, using a highly redundant placement and exploiting machine-learning to understand sensor outputs. Integrating sensing directly into the actuation chamber has also been tested

by multiple groups [71, 144]. This reduces the overall complexity of the whole system, but to the detriment of sensing accuracy, and the ability to target sensing to key areas.

### **Data Interpretation and Analysis**

Once the sensors have been placed, interpreting and analysing the data is another challenge. With the high dependency of sensing on the soft body, the characterisation of the sensor on its own, or the sensor in the system, can make a huge difference. The characterisation of the sensor *ex situ*, can allow easy comparison between the performance of different sensors, although it does lead to possible changes when placed on the robot. In contrast, testing directly on the system allows for the behaviour to be understood, but gives no indication of behaviour in other systems.

Another key aspect of interpretability are the complex behaviours of soft materials, which also affect the sensors made from them. As such, soft sensors are prone to hysteresis behaviours, overshoot, as well as nonlinearities in signals [173]. Some research has been done to try to ease interpretability, without compromising on softness, by combining ionic solution interfacing with liquid metal, to achieve a soft sensor which decouples the signal routing and the sensor, and increases its gauge factor [30], but involves more complex fabrication.

### **Biocompatibility**

Finally, with the use of soft sensors in biomedical applications in mind, the use of biocompatible materials, as well as the durability of sensors through wear and tear is essential. In terms of resistive sensors, one key medium being used, which is popular for medical application, is ionic solutions. A sodium chloride solution has been proposed for the integration in a flexible surgical tool [160], where gold sputtering and glycerol were proposed as solutions to the issues of corrosion and air bubbles respectively. Sensors using ionic solutions have also been proposed in wearable applications [206]. In this case, the sensor showed good sensitivity and was applied in cases to detect bending of joints, as well as the finer motions of breathing, speaking and swallowing.

## **2.7 Conclusion**

To conclude, this literature review has shown examples of robots in the medical field before substantiating the need for softer technology. From there, the field of soft robotics was introduced and the need for reliable actuators, accurate sensors, and smart control was shown to be essential, especially with regards to safety-critical medical systems. A variety of soft

actuators were presented, and highlighted the challenge of overcoming the theoretical infinite degrees of freedom and highly nonlinear behaviours. Soft robotic control and modelling strategies were then presented, ranging from the application of conventional control strategies, to more data-driven approaches. The purpose and demand for soft sensors, to provide vital information about the robot itself and the environment and enable closed-loop control, was underlined and explored in the final section. The complicated characterisation and difficulty of interpretability of soft sensors also brought to the foreground the need for new ways of thinking about soft sensors. The culmination of the soft robotics review, including the soft control and sensors, alongside the specific requirements drawn from the medical context, has made clear the need for space efficiency and the advancement of soft sensors to establish safer soft implantable robots. This section ends with questions around the research gaps that still need to be addressed.

## 2.8 Research Questions

Given the current state-of-the-art, the questions that this research seeks to answer are:

1. What opportunities are there to go beyond the current sensing frameworks to develop smart/intelligent multipurpose sensors that can seamlessly blend into a range of soft medical robots and provide added information on the state of the robot itself or the biological surroundings?
2. How can we maximise the information available to us about our medical robot and the environment its acting in, whilst being the most space-efficient possible, to fit into the tightly constrained gastrointestinal tract?
3. How can we make soft and safe implantable robots, ensuring compliance and biocompatibility of the robot, without foregoing on sensed information, vital for providing effective therapies?



## Chapter 3

# Soft Force and Displacement Detecting Fluidic Sensor-Actuator

### 3.1 Preface

In this first technical chapter, the work from [81] and its extension in [82] is presented. Fluidic actuation is integrated with displacement and pressure sensing to achieve a compact and versatile soft fluidic ballooning sensor-actuator (SA), shown in several applications (Fig. 3.1). A comprehensive study of three different diameters SAs is shown under both static and dynamic conditions and under both loading and constraint, as well as for angled loading under static characterisation. The SA is also shown in two different applications: a soft and a rigid-bodied probe example capable of distinguishing objects and regions of varying stiffness; and a soft robotic implant, tested in *ex vivo* tissue and able to predict force and displacement, key for future estimations of tissue lengthening and stiffness. The contributions of this chapter include: (1) the design and fabrication of a soft fluidic ballooning SA; (2) the characterisation of the versatile SA under load and constraint for both static and dynamic conditions; (3) the demonstration of the SA in both two probe examples, able to distinguish between soft and stiff objects or regions, and (4) the testing of the SA as part of a soft robotic implant, that can predict forces and displacement.

The work is based on ©2022 IEEE. Reprinted, with permission, from [81]. Including its further extension submitted as [82].

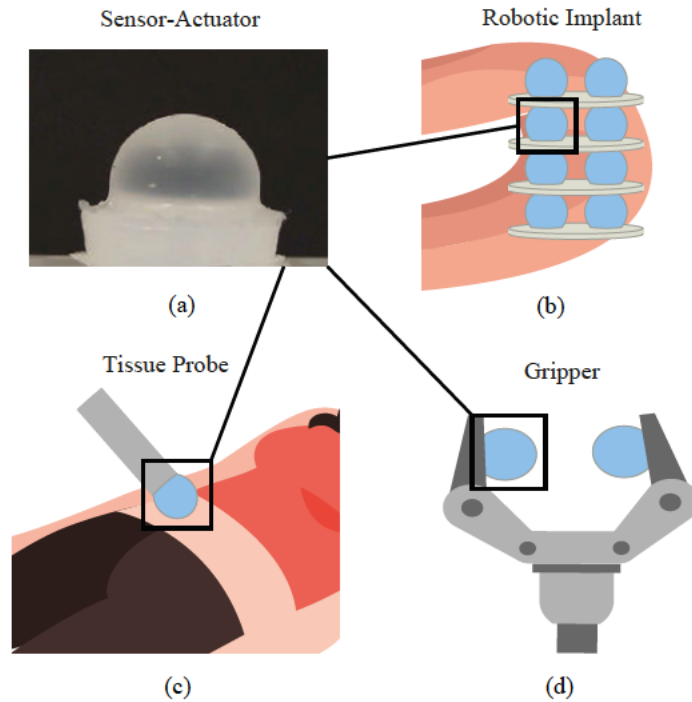


Fig. 3.1 Application examples of the SA. (a) Image of the SA. (b) Schematic of the SA as a ballooning robotic implant to induce tubular tissue growth [72]. (c) Schematic of the SA in a probe example to palpate the abdomen. (d) Schematic of the SA in a gripper example used to detect and grasp objects.

## 3.2 Introduction

In biomedical robotics, safety and effectiveness are key desirable characteristics. In terms of safety, sensing is vital to provide feedback on the state of the robot itself, as well as its surroundings, and therefore allow medical professionals to make informed and safe therapeutic decisions. The choice of material is equally important for safety though, and soft robots have proven to be naturally adept in such situations. Their inherent compliance enables them to conform to their surroundings and allows for safe interaction with the human body, although it does in turn present challenges for actuation and sensing that need to be overcome [32]. In addition, to be effective, the actuation method of the robot is vital. The robots need to be capable of both exerting required forces and long extension, to induce the necessary changes in the human body, as well as to have the ability to grow with tissues if necessary [138]. Finally, all of these features need to be accomplished in a compact form, to not hinder the natural movement of the body and enable the robot to work within the confined space of the human body.



In particular, two applications that could benefit from a versatile soft sensor-actuator are palpation and tissue lengthening. For palpation, there are already examples of robotic probes used on the abdomen [73], to identify areas with abnormal nodules, as well as soft tactile sensors to be used laparoscopically [20] and endoscopically [210] to identify solid tumors. However there is still scope to minimise the invasiveness of these tools, as well as to develop a more modular, soft and easily integrated sensing tool, using tactile sensing and dynamic inflation to interactively diagnose tissues and map stiffnesses. For tissue lengthening, which is a potential treatment for short bowel syndrome, the current developed solutions using spring-mediated distraction enterogenesis [146][145], rigid telescopic expansion actuators [45][148] or shape-memory polymers [55] fail to provide large, controllable, patient-specific and ultimately safe lengthening of tissues. This leaves a gap for new solutions, using softer approaches to more closely match the tissue to be lengthened and surrounding tissues, ensuring the device can be placed intraluminally using minimally invasive approaches, as well as embedding sensing into the devices to allow more in situ and informed decisions and targeted therapies. A soft sensor-actuator could therefore be a key building block to tailor robotic implants to the patient's lengthening needs, as well as allowing surgeons to monitor and amend the mechanical distraction exerted on the tissue to allow for optimal tissue growth.

In both applications, distinguishing between external and internal stimuli is key in ensuring the effectiveness and safety of the technologies. Many different strategies for sensor interpretation of soft sensors have previously been explored, from varying sensor elements, to strategic placement of sensing elements, as well as exploiting the sensor geometry itself [190]. Combining different sensors over the body of the robot, can both ease the interpretability of the sensing and allow for better distinction between stimuli. However, having a multitude of sensing modalities often increases the overall complexity of the system and usually requires more space. Another strategy is placing simple sensors in either targeted regions [209] or targeted directions [132] or randomly distributing sensors [178], all of which have proven to be effective, but similarly come at an obvious cost of space. Finally, sensor geometry has previously been exploited to differentiate between internal and external stimuli [62][125], but is application dependent and often requires additional space.

Among soft robots, fluidically-driven actuators have proved their abilities to exert forces and deformations with potential for biomedical applications [51][75]. The sub-category of ballooning membrane actuators is additionally considerably compact, being capable of generating a large growth from a very small footprint [72][88]. Pneumatic actuators are subject to snap-through instabilities though [126][129], which can lead to uncontrolled expansions and ultimately bursts. In contrast, hydraulic actuators can be volume controlled,

because of the incompressibility of liquids, and as such do not experience such instabilities. These hydraulically ballooning membrane actuators, therefore strike a balance between large growth, high load-bearing capabilities and improved behaviour predictability.

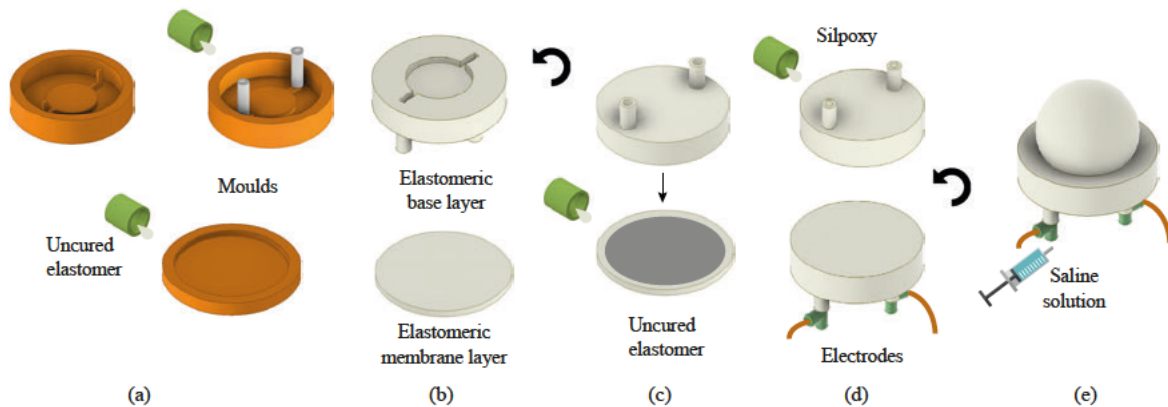


Fig. 3.2 Schematic of the fabrication procedure. (a) Tubing is pre-inserted into the mould and uncured Ecoflex 00-50 silicone is poured into the moulds. (b) The room-temperature cured parts are removed from the moulds and (c) bonded with uncured Ecoflex 00-50 and left to cure. (d) (Top) The tubing is reinforced with Silpoxy silicone glue and (Bottom) electrodes are connected into T-tube connectors. (e) Finally, the SA is filled and inflated with a saline solution.

Given the importance of space efficiency, combining sensing and actuation into a single module is an optimal solution to fully utilise the amount of space available, whilst also allowing in situ sensing. Examples of soft actuators capable of sensing and actuation simultaneously include dielectric membranes [150], and magnets [162][56]. Work has also already been undertaken on adding sensing capabilities to fluidically actuated robots, although this research has to date been focused on specialised actuators [71][144] rather than singular elements that can be integrated in a variety of robotic applications. There has been some research on ballooning membranes and modeling the pressure inside ballooning membranes, to understand the forces applied and recreate the shape during inflation [54][99], however focuses on pneumatic actuation. In addition, there is still a lack of understanding in the change in geometry of a membrane actuator when it is constrained and real-time tracking of growth still needs to be achieved. This leaves a clear gap to combine sensing elements to expand on the sensing information available for ballooning membrane actuators and transform them into universal building blocks for biomedical robots.

## 3.3 Materials and Methods

### 3.3.1 Design

The principle of operation of the soft sensor-actuator (SA) relies on combining pressure and resistance sensing, in order to complement the overall picture the pressure gives on the loading on the sensor with the resistance, which provides insights into the cross-sectional areas across the SA between the electrodes. By combining sensing elements, the study on this SA also differentiates itself from other studies on membrane actuators which focused on pressure sensing only. The choice of a membrane actuator, makes the SA universal and applicable to many soft robots, in addition to separating itself from the other studies of using conductive fluids in soft actuators which were based on Pneunet [144] and flexible fluidic actuators [71]. The mix of sensing and actuation makes it possible for the SA to be used in actuation and sensing situations separately, as well as combining the functionalities to more dynamically sense the surrounding environment. Finally, given the implantable application in mind, the SA's diameter was also limited to 20 mm to fit inside the narrowest organ lumen in the gastrointestinal tract.

### 3.3.2 Fabrication of the soft SA

The soft SA is made using a layered approach (Fig. 3.2): first casting silicone layers, and then bonding them with uncured material. The silicone layers are cast out of Ecoflex 00-50 (Smooth On Inc) using PLA moulds printed on a 3D printer (Prusa i3 Mk3). The Ecoflex 00-50 is mixed using a mechanical mixer (ARE-249 Mixer, Thinky), and degassed for three minutes, before being poured into the moulds with the pre-inserted tubing. Any excess Ecoflex 00-50 in the membrane mould was scraped off to help keep consistency in thickness, with a 1 mm membrane chosen as a good compromise between manufacturing error tolerance, expansion achievable and structural strength when inflated. Once poured and leveled, both parts were left to cure at room temperature. Once cured, the elastomeric base channel layer is bonded to the top ballooning membrane layer using a thin layer of uncured Ecoflex 00-50 scraped over the membrane, and again left to cure at room temperature. The tubing is then reinforced with Silpoxy (Smooth-On Inc) and connected to a T-tube, also connected to the syringe and an electrode, made by stripping 5 mm of the plastic covering at the end of a tinned copper wire, for easy integration of resistive sensing. Finally, the SA is filled with a biocompatible 0.9% saline solution (9g salt for 1L distilled water), with continuous inflill and monitoring enabled by the T-tube connectors. The finished SA is 20 mm in diameter and 5 mm tall, chosen for its size well within the inner diameters of the different organs within

gastrointestinal tract, and is capable of up to 10 mm of pure extension above its original height with 2 mL of liquid.

### 3.3.3 Experimental Setup for SA Characterisation

To test the SA, a combination of custom-built rigs are used (Fig. 3.3). These rigs apply loads and constraints, and input different volumes to the SA, whilst monitoring its pressure and resistance.

Firstly, to control the inlet and outlet volumes of the SA, two syringe drivers are used. The syringe drivers are controlled with an Arduino-Uno and a CNC shield (Youmile), allowing full flow through the SA, and easing air bubble removal. Secondly, the resistance of the 0.9% saline solution across the SA is calculated using a voltage divider circuit. A 17 k $\Omega$  resistor, connected in series with the SA, is connected to a benchtop function generator (Gwinstek SFG-1013), that could in the future be replaced with a portable custom-built PCB, supplying a 1 V 1 kHz voltage sine wave. A high frequency AC signal was chosen to prevent electrolysis of the saline solution and the series resistor value was selected to be close to the measured resistances for good sensitivity over the sensing range. The series resistor is monitored using an NI USB-6009 DAQ board and LabVIEW at a frequency of 12 kHz. Thirdly, the pressure in the SA is measured using a pressure transducer throughout the experiments. Finally, the testing platform shown in Fig. 3.3(b) is used to apply loads to the SA uniformly, by adding weights onto the 20g sliding plate, guided using ball bearings press fit into the 3D-printed sliding plate and two linear rods. To constrain the growth of the SA, endstops are added to the rods, and stop the displacement of the sliding plate at a given height.

### 3.3.4 Experimental Protocols

The soft SA was characterised both statically and dynamically, to investigate its use under pure static sensing conditions, as well as for dynamic sensorised actuation. Additionally both bubble height and loading were tested, as key information to estimate tissue lengthening and tissue stiffness for the applications in mind. SAs with expanding area diameters of 8 mm, 10 mm and 12 mm, chosen to test varying dimensions while still allowing a thick enough side wall to limit unwanted sideways expansion, were tested for the following experiments.

#### Static Characterisation

For the static characterisation, three experiments were conducted, to validate the sensing capability for force and displacement over uniform ranges of loads and constraint heights, as well as angled loading.

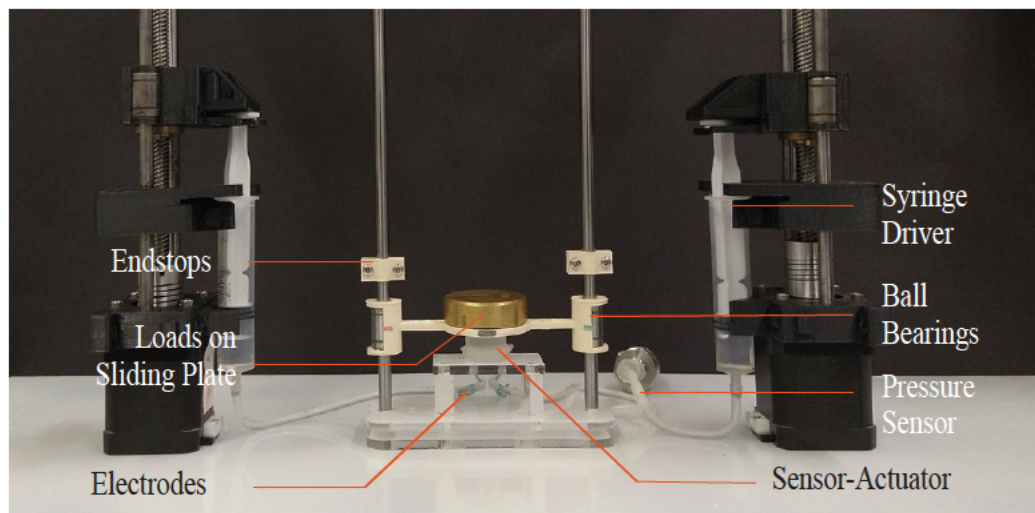
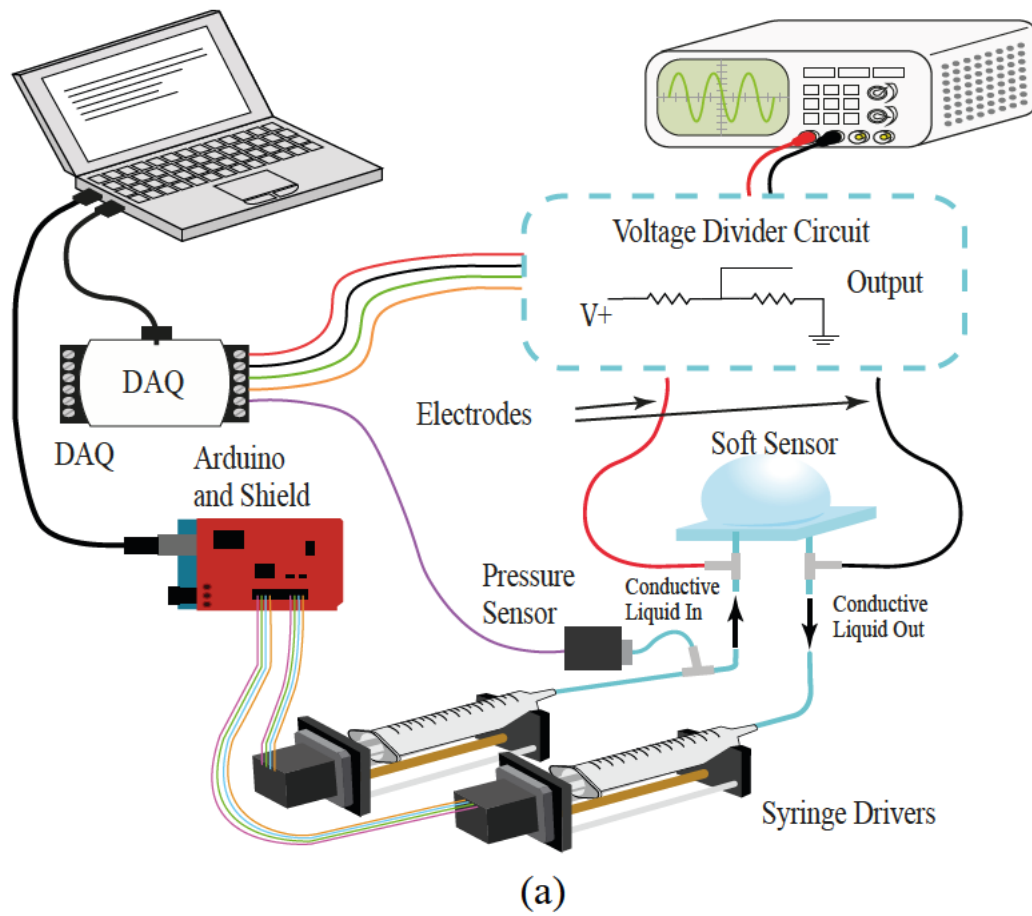


Fig. 3.3 Experimental setup. (a) Schematic of the electronic setup for inflation of the SA with automatic syringes and the sensing conditioning circuit. (b) Front view photo of the characterisation setup for an SA.

For the first static characterisation, the aim was to understand the correlation between the height of the SA after compression and its resistance and pressure. The height of the SA was measured from the base of the ballooning surface to the top of the bubble or constraining plate, equivalent to the height  $h_a + h_b$  in Fig. 3.4. To achieve this, the SA was again filled with the three different volumes of 1, 1.5 and 2 mL, conservatively chosen for good inflation while reliably not bursting. It was then progressively compressed in 1 mm increments up to 5 mm. The initial height of the SA was determined using experimental photos and an image-tracking software (imageJ).

The second experiment aimed to understand the relationship between an even load applied and both the resistance and pressure. For this experiment, the SA was inflated to 1, 1.5 and 2 mL, and loaded vertically with calibrated weights from 0 to 210 g in 30 g increments. The maximum load of 210 g was chosen for its clinical relevance in a biomedical robotic implant for tissue growth needing to exert forces up to 2 N (Fig. 3.1(b))[40].

Finally, the third static characterisation was designed to observe the variations in both resistance and pressure, when the SA is subject to angled loading. For this, the SA was inflated with 1 mL of saline solution and loaded with calibrated weights from 0 to 210 g in 30 g increments. The loading plate was angled at 15°, 30° and 45° and tested in two different orientations, to determine the effect of both loading angle and loading orientation with respect to the electrodes. Firstly for the accentuated angled loading orientation, the load was more accentuated towards one of the electrodes, with the contact surface plane angled with respect to the bisecting line of the two electrodes. Secondly for the homogeneous angled loading orientation, the load was homogeneously acting across the length of the bubble SA, with the contact surface plane angled with respect to the line between the two electrodes. For all static testing, the experiments were repeated five times for each condition, with resistance and pressure being monitored.

The static characterisation results of the constrained and loaded experiments were also used to develop models to predict the load and the height of the bubble in the application experiments. For this, linear regression models with intercepts were determined using MATLAB's in-built `fitlm` and `regress` function using the raw data of the volume, the load or the constraint applied and a combination of the measured pressure and resistance. With the exception of the static constrained data, where the log of the pressure data was used instead of the raw data, as a nonlinear behaviour was observed.

### Dynamic Characterisation

For the dynamic characterisation, the SA was again tested under loaded and constrained conditions. The objective of these experiments was to test the SA under dynamic actuation

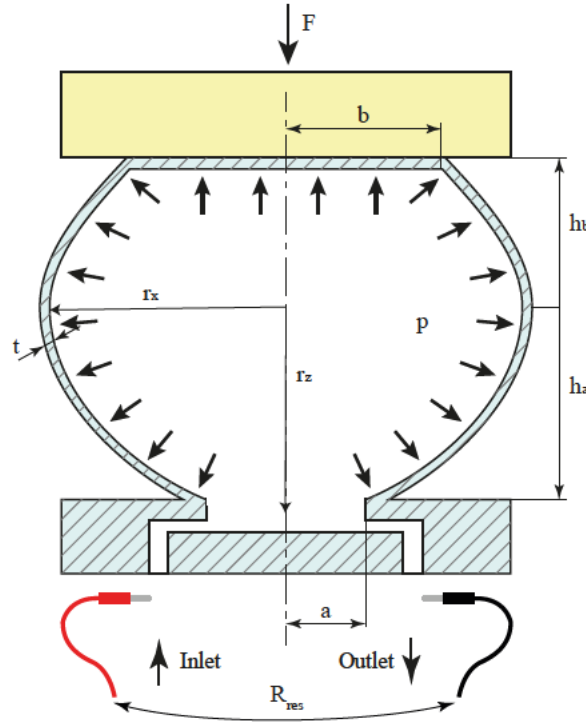


Fig. 3.4 Schematic of the inflated SA.

and understand the effect on the resistance and pressure when constrained or loaded, for possible applications in a tissue growing robotic implant or for dynamic sensing in a probe or gripper application (Fig. 3.1). For the constrained condition, the SA was constrained at 4, 6 and 8 mm above the top of the uninflated SA, mimicking objects constraining its growth. The SA was then inflated with 2 mL of solution in 0.5 mL increments, with pressure and resistance being monitored. For the loaded condition, the SA was pre-loaded with calibrated weights from 0 to 210 g, again given the clinical relevance, and then filled with 2 mL of solution in 0.5 mL increments, with pressure and resistance being monitored.

### 3.4 Modelling

In the following section, the analytical model for the pressure and resistance for the SA is described with a number of assumptions about its shape being made (Fig. 3.4). The work is extended from previous work [72] with the additional hydraulic consideration and provides some insights of the impacts of various parameters, but the full modeling would be the scope of a different paper.

For the model, the initial radius, thickness and volume of the ballooning membrane  $a$ ,  $t_i$  and  $V_m$  respectively, and the Young's modulus  $E$  of the elastomer are the only constant

parameters. All the other variables change as the membrane is inflated and the geometry of the bubble changes.

To begin, assuming a spheroidal frustrum shape of the SA throughout the inflation and additional loading or constraining, as well as the incompressibility of the saline solution, the internal volume of the SA can be defined as:

$$V_f = \frac{4}{3}\pi r_x^2 r_z - \pi ab \left( \frac{4r_z}{3} - h_a - h_b + \frac{h_a^3}{3r_z^2} + \frac{h_b^3}{3r_z^2} \right), \quad (3.1)$$

where  $V_f$  is the volume of the fluid,  $r_x$  and  $r_z$  are the radii of the spheroid,  $h_a$  and  $h_b$  are the heights from the center to the base and object respectively,  $a$  is the initial radius of ballooning membrane of the SA and  $b$  is the radius of the top of the bubble.

Additionally, assuming the membrane is incompressible, the final thickness  $t$  can be related to the initial thickness  $t_i$  by:

$$V_m = \pi a^2 t_i = S_m t, \quad (3.2)$$

where  $V_m$  is the volume of the membrane, and  $S_m$  is the surface area of the spheroidal membrane, defined as:

$$S_m = 2\pi a \left( \int_0^{\text{atan}\left(\frac{r_x h_a}{r_z a}\right)} \sqrt{r_x^2 \cos^2 \theta + r_z^2 \sin^2 \theta} d\theta + \int_0^{\text{atan}\left(\frac{r_x h_b}{r_z b}\right)} \sqrt{r_x^2 \cos^2 \theta + r_z^2 \sin^2 \theta} d\theta \right), \quad (3.3)$$

Using shell theory, the stress can be found using:

$$\sigma = \frac{p A_{\text{cross}}}{t l_m}, \quad (3.4)$$

where  $\sigma$  is the stress,  $p$  is the pressure,  $A_{\text{cross}}$  is the cross-sectional area ( $A_{\text{cross}} = \pi b^2$ ) and  $l_m$  is the circumference of the bubble at the mid-cross section ( $l_m = 2\pi b$ ).

Assuming a neo-Hookean model for the Ecoflex 00-50 membrane and an equibiaxial load, the stress can be defined as follows:

$$\sigma = \frac{2E}{3} \left( \left( \frac{t}{t_i} \right)^2 - \left( \frac{t_i}{t} \right)^4 \right), \quad (3.5)$$

where  $E$  is the Young's modulus.



The force  $F$  acting on the SA can be written as:

$$F = p\pi b^2. \quad (3.6)$$

Using Pouillet's law, the resistance  $R_{res}$  across the SA can be described as:

$$R_{res} = \rho \int_{-r}^r \frac{1}{A(x)} dx, \quad (3.7)$$

where  $\rho$  is the resistivity of the saline solution and  $A(x)$  is the varying cross-sectional area of the bubble across its diameter, assuming flat and centered deformation under loads. The varying cross-sectional area  $A(x)$  can be written as:

$$A(x) = \int_{\max(-h_a, -r'_z)}^{\min(h_b, r'_z)} 2\sqrt{r_x'^2 \left(1 - \frac{z^2}{r_z'^2}\right)} dz \quad (3.8)$$

where  $r'_x$  and  $r'_z$  are the changing diameters of the varying elliptical segment, which can be obtained from the standard equations of an ellipse and a spheroid.

Finally, using the theory of minimum potential energy, the model becomes an optimisation problem of minimizing energy  $E_p$ , written as:

$$E_p = \int_{V_m} W dV - \int_{V_f} p dV - mg(h_a + h_b) \quad (3.9)$$

where  $E_p$  is the sum of the elastic energy of the membrane, with  $W$  the work of the hyperelastic membrane defined by the neo-Hookean material assumption, the negative energy from work of the pressure by the fluid and the negative energy from the external loading.

There is no analytical solution to the equations above given the high complexity and interdependence of terms, however, accuracy is lost with further simplification. Nonetheless some qualitative findings can be derived from the equations: thicker membranes require more pressure to inflate, smaller diameter membranes require more pressure to inflate and heavier loads flatten the bubble more for a given pressure.

## 3.5 Results and Discussion

### 3.5.1 SA's Static Characterisation

The results from the static constrained characterisation are shown in Fig. 3.5. Across all volumes and for all diameters of SAs, the pressure decreases as the bubble height increases,

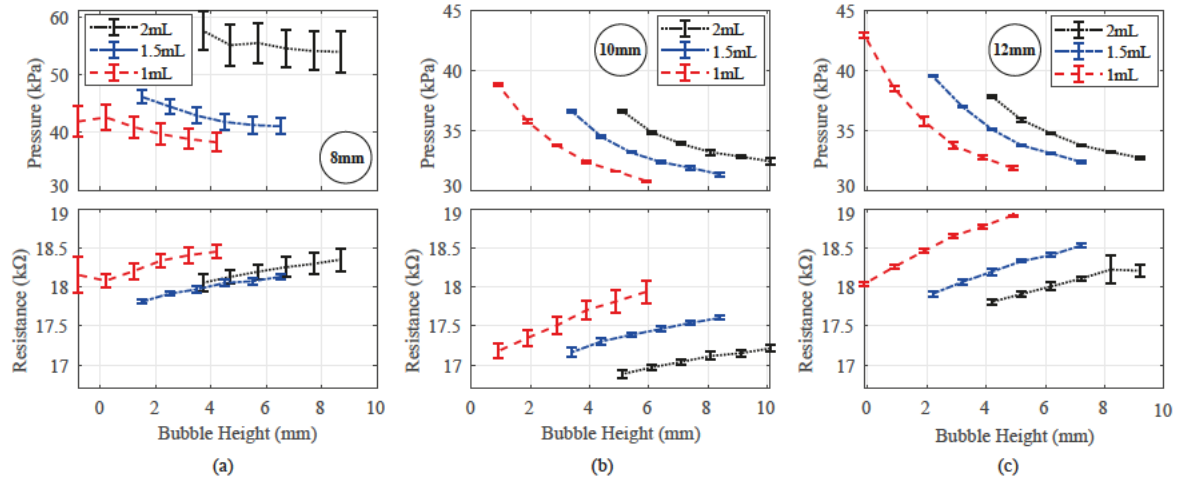


Fig. 3.5 Static characterisation of the SA under constraint. (a) (Top) Pressure and (Bottom) resistance with respect to the bubble height (height between the base of the ballooning surface to the top of the bubble or constraining plate) for the 8 mm, (b) 10 mm and (c) 12 mm diameter SAs. The SA was initially inflated to 1 mL (red), 1.5 mL (blue) and 2 mL (black). The SA was then progressively compressed in 1mm increments up to 5 mm, and the bubble height after compression was measured. The error bars show the standard deviation across five trials.

equivalent to less compression. The bubble height, the height between the base of the ballooning surface and the top of the bubble or constraining plate, is inversely proportional to pressure and follows a good logarithmic trend, reflected by the  $R^2$  value of 0.85 and 0.87 for the 10 and 12 mm diameter SAs respectively, but the 8 mm diameter SA experiences large errors in pressure, potentially due to off-centered loading from the manual loading protocol in the static characterisation, and accentuated further on the sensor-actuators with smaller diameters that have a more rounded inflation, which is reflected by the poor  $R^2$  value of 0.20 (Table. 3.2). The amplitude of the difference in pressure also decreases for larger volumes, with the 1 mL having the largest difference in amplitude. In contrast, the relationship between height and resistance is linear, but again does show a flattening of the curve for larger volumes. The linear relationship is again reflected in the  $R^2$  values. For the 12 mm diameter case, the multiple linear regression using resistance only shows a slight improvement from the pressure only model, for the 10 mm, the results are worse due to the larger variation in the 1 mL data, but for the 8 mm diameter it shows large improvement as the errors for the resistance measurements are relatively smaller.

The results from the loaded static characterisation are shown in Fig. 3.6. Across all volumes and for all diameters of SA, the relationship is proportional between load and pressure. The good linear correlation is reflected by the  $R^2$  value of 0.95 for 10 and 12 mm

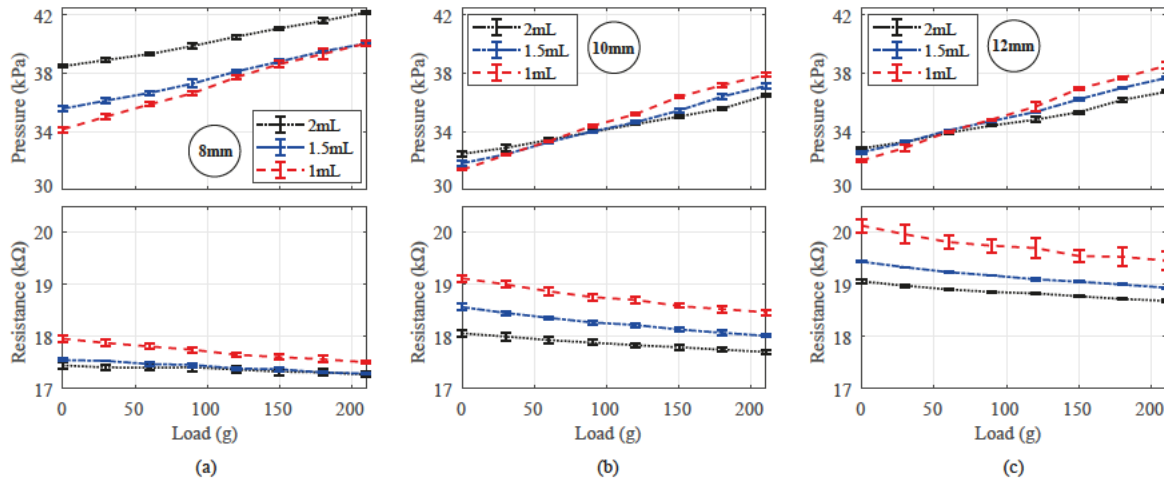


Fig. 3.6 Static characterisation of the SA under loading. (a) (Top) Pressure and (Bottom) resistance with respect to load for the 8 mm, (b) 10 mm and (c) 12 mm diameter SAs. The SA was initially inflated to 1 mL (red), 1.5 mL (blue) and 2 mL (black). The SA was then progressively loaded with weights from 0 to 210 g in 30 g increments. The error bars show the standard deviation across five trials.

Table 3.1 Multiple linear regression  $R^2$  value for models with pressure readings (P), resistance readings (R) and combined readings (P&R), in combination with the diameter information for constrained static characterisations.

| Static characterisation under constraint |        |        |        |
|------------------------------------------|--------|--------|--------|
|                                          | P      | R      | P&R    |
| 8mm                                      | 0.1989 | 0.3415 | 0.5335 |
| 10mm                                     | 0.8497 | 0.7334 | 0.8513 |
| 12mm                                     | 0.8693 | 0.8803 | 0.8928 |

diameters and 0.87, for 8 mm diameter for the multiple linear regression using pressure only (Table. 3.2). Additionally, the pressure for larger volumes is initially greater, but the gradient is less steep, as the load is better distributed. In contrast, the relationship between load and resistance is inversely proportional, but overlaps across neighboring loads and adds little information to the curve fitting. The resistance is also shifted upwards for greater volumes, in accordance with Pouillet's Law dictating that resistance is inversely proportional to the cross-sectional area, hence decreasing the resistance for larger volumes.

Fig. 3.7 shows the results from the static characterisation under angled loading. Irrespective of the angle of loading, as the load is increased, the difference in pressure and in resistance relative to the unloaded state increase and decrease respectively. Additionally, as the loading angle is increased, the difference in pressure is smaller, with the smallest

Table 3.2 Multiple linear regression  $R^2$  value for models with pressure readings (P), resistance readings (R) and combined readings (P&R), in combination with the diameter information for loaded static characterisations.

| Static characterisation under loading |        |        |        |
|---------------------------------------|--------|--------|--------|
|                                       | P      | R      | P&R    |
| 8mm                                   | 0.8879 | 0.5952 | 0.9226 |
| 10mm                                  | 0.9501 | 0.8565 | 0.9501 |
| 12mm                                  | 0.9482 | 0.6711 | 0.9489 |

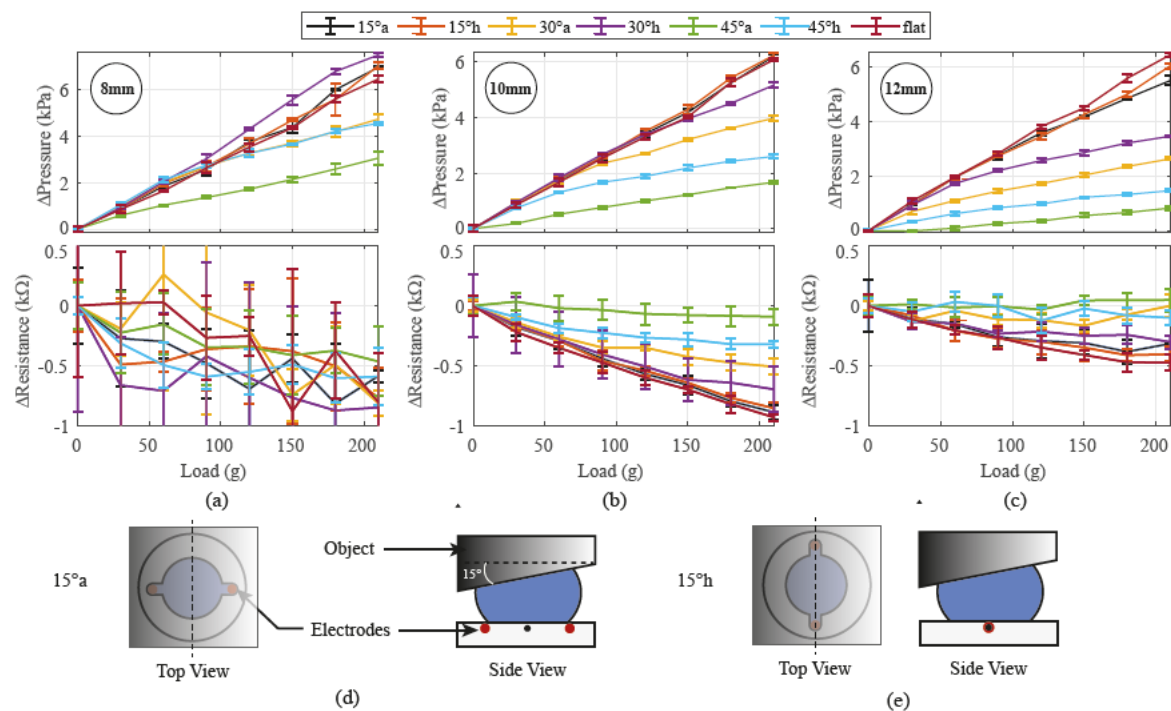


Fig. 3.7 Static characterisation of the SA under angled loading. (a) (Top) Pressure and (Bottom) resistance with respect to load for 8 mm, (b) 10 mm and (c) 12 mm diameter SAs. Illustration of the angled object with respect to the electrodes for (d) a 15° accentuated indentation (15°a) and (e) a 15° homogeneous indentation (15°h). The sensor actuator was initially inflated to 1 mL and then loaded with angled platforms of 15°, 30°, and 45° in two different orientations, as well as a flat object for reference. The SA was also progressively loaded with weights from 0 to 210 g in 30 g increments. The error bars show the standard deviation across five trials.

angle loaded conditions being almost equal to the flat platform condition. This is likely from the decreased normal force for increased loading angles, with little or no variations in the membrane surface contact area, requiring less pressure to maintain force equilibrium at the contact surface of the membrane (Eq.6). The same is true for the resistance, but

on the negative scale, although for the 8 mm diameter SA, the difference in resistance is majorly affected by errors, and for the 12 mm diameter SA, the difference in resistance is much smaller. The orientation of the load also has an effect on both pressure and resistance. When the loading is more accentuated towards one electrode, the difference in pressure and resistance is less than when the loading is more homogeneous. The difference in pressure could occur from the fact that the pressure sensor is on the side nearest the most indented region, but the difference in resistance is expected from the difference in the cross-sectional shape along the SA that the different loading orientations affects.

Across all the static characterisations, there is mostly a good separation between the different conditions, allowing for good interpretability, as reflected by the high  $R^2$  values, with the 10 mm diameter SA showing the best performance relative to the others. Under loading, the 10 mm diameter SA also had the best pressure sensitivity for all three volumes, peaking at an average 0.93 kPa per 30 g when filled with 1 mL, and had similar resistance sensitivities as the 12 mm SA of just less than 0.1 k $\Omega$  per 30 g. Under constraint, the 10 and 12 mm SAs showed comparable behaviours with peak pressure sensitivities of 1.6 and 2.2 kPa per mm and resistance sensitivities of  $-0.15$  and  $-0.18$  k $\Omega$  per mm respectively. The estimated resolutions across all SAs was improved using pressure, with the best static loading resolution of 3.7 g achieved for the 12 mm sensor filled with 1.5 mL, compared to the resolution of 8.4 g for the same conditions using resistance sensing. For the static constraint conditions, similarly the 10 and 12 mm SAs performed best, with the best resolution of 0.07 mm achieved using pressure sensing in the 12 mm SA, over twice as good as using resistance sensing where the best resolution achieved for the 12 mm SA was 0.17 mm. The predictions of load, from the  $R^2$  values, are better than those of height in most cases though, with the exception of the 8 mm diameter SA under constraint. However, as the volume increases, there is some loss of information about the height and load, as the curves flatten and hence the resolution of resistance and pressure between test conditions becomes smaller and harder to differentiate. Between the static characterisations there is also a slight drift in resistance measurements, as a result of the change in conductivity of the saline solution with respect to temperature. This change in resistance, would have less of an impact in highly temperature-controlled environments, such as the targeted human body, but is worth considering for long-term applications in less controlled environments. This difference could be calibrated though by using a minimally filled actuator and establishing a base resistance. Additionally, the resistance measurements have the advantage of being better isolated from the environment by exploiting the difference in conductivity of the ionic solution compared to the wire. The pressure measurements on the other hand are susceptible to artifacts from loading across longer tubing and from delayed measurements of changes in pressure, but are



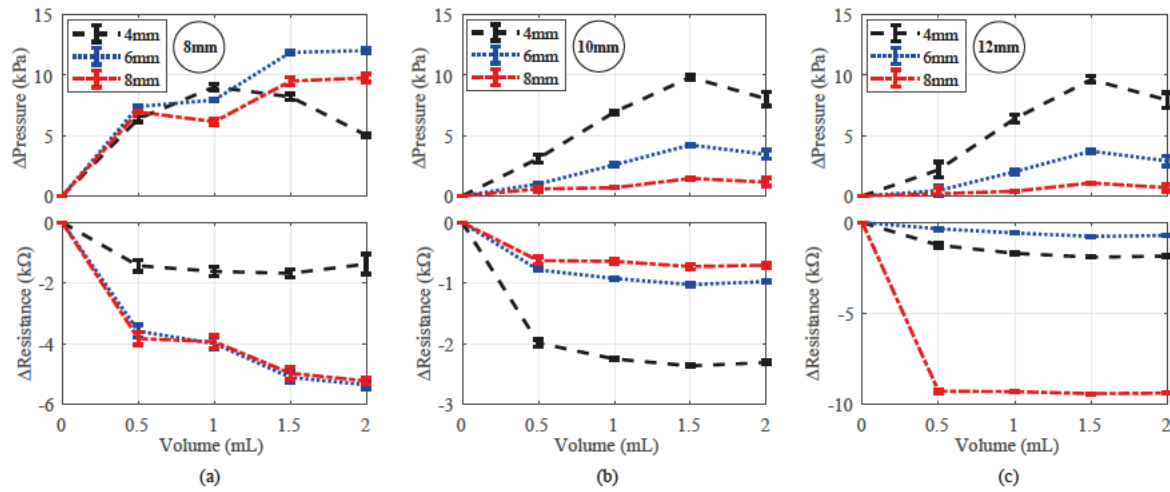


Fig. 3.8 Dynamic characterisation of the SA under constraint. (a) (Top) Difference in pressure and (bottom) difference in resistance relative to an unloaded SA for various constraints during volume increase for the 8 mm, (b) 10 mm and (c) 12 mm diameter SA. The SA constrained at 4, 6 and 8 mm above the top of the uninflated SA, and was injected with 2 mL in volume in 0.5 mL increments. The error bars show the standard deviation across five trials.

immune to changes in temperature and are consistent across testing conditions. Overall, with the current design, fabrication and testing conditions the 8 mm diameter SA demonstrated a less consistent behaviour with a poorer sensing measurement range, so unless a smaller SA bubble diameter is needed, when scaling for an application, it is not the ideal choice. The 10 and 12 mm diameter SAs exhibited similar performances, with the 12 mm diameter SA having the best resolution. However, if inflated bubble height is a consideration, the 10 mm diameter SA achieved greater heights for the same amount of liquid and the SAs filled with 1.5 mL of liquid proved to be the most consistent for all diameters of SA. Considering the above characterisations, as well as specifications desired for *in vivo* applications, e.g., extension capacity, compactness, behaviour predictability, the 10 mm sensor was chosen for the demonstrations in this paper.

### 3.5.2 SA's Dynamic Characterisation

The dynamic characterisation for the constrained SA is shown in Fig. 3.8. The difference in pressure and resistance, with respect to an unloaded SA are shown as the volume is increased in 0.5 mL increments up to 2 mL. The largest difference in pressure is observed for the 4 mm constraint condition for both the 10 and 12 mm diameter SAs, but, for the 8 diameter SA, the pressure is initially highest for the lowest constraint before it converges back toward 0. The difference in pressure also starts decreasing at 1.5 mL for most of the constraint conditions,

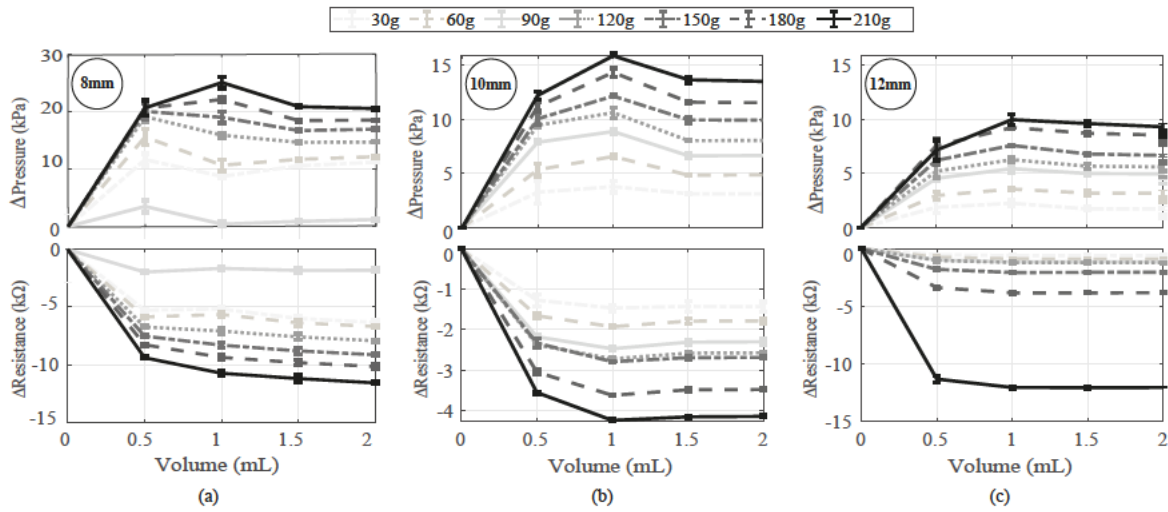


Fig. 3.9 Dynamic characterisation of the SA under loading. (a) (Top) Difference in pressure and (bottom) difference in resistance relative to an unloaded SA for various loads during volume increase for the 8 mm, (b) 10 mm and (c) 12 mm diameter SA. The SA was loaded from 30 to 210g in 30g increments and injected with 2 mL in volume in 0.5 mL increments. The error bars show the standard deviation across five trials.

which corresponds to when the SA starts to balloon more radially. In contrast, the difference in resistance decreases, as additional volume is injected. The difference also tends to be larger, as the constraint height decreases, with a few exceptions. Additionally, there is also a slight increase in resistance, matching the decrease in pressure for 2 mL, when the radial ballooning leads to a lengthening of the SA.

Fig. 3.9 shows the results from the loaded dynamic characterisation. The difference in pressure and resistance, with respect to an unloaded SA are shown as the volume is increased in 0.5 mL increments up to 2 mL. For all loads, the pressure is greater than the unloaded actuator, and the difference is greatest for larger loads, but maintains good separation between loads. For all diameters of SA, the difference in pressure peaks at 1 mL, before converging. The convergence is most marked for the smaller diameter SA, where the SA has a more rounded and less flat shape during inflation and experiences the most deformation when loaded. In contrast, the resistance decreases compared to the unloaded actuator, but, again, the greatest difference is for the larger loads, as the shape of the actuator becomes flatter and the cross-sectional area more uniform. The difference in resistance is also mostly smaller as the diameter size increases, as it experiences less shape change.

Across the dynamic characterisation, the separation between conditions is still good, demonstrating the ability of the SA to both sense and actuate. However again, as the volume increases, there is a loss of information about the height and the loads, due to the curves

converging, from the radial ballooning and lengthening of the actuator at larger volumes. This imposes a trade-off for actuators, if both large growth and accurate sensing are needed, such as for a robotic implant. In such cases, having dedicated sensing layers or exploiting the active dynamic behaviour could provide an answer. In addition, the change in shape of the actuator and subsequent uncharacteristic changes in pressure and resistance for larger volumes, could be improved with a more rigid base layer. This would prevent the lengthening of the actuator and prevent the increase in resistance. Furthermore, to improve the distinction between pressure readings for varying loads, smaller volumes would need to be used, or distributing heavier loads across several SAs could minimise the radial ballooning and prevent the observed saturation of the readings.

In conclusion, input volumes up to 1.5 mL lead to the best differentiable predictions of height and load. The smaller the input volume, the greater the resolution of pressure and resistance across the tested heights and loads, as the bubble of the SA experiences the greatest shape difference, without affecting its base. Additionally, all diameters of SA have a similar height when unloaded and unconstrained, but the 12 mm diameter SA has better load-bearing capabilities from an overall flatter shape, reflected by the slower saturation in pressure and resistance for larger volumes. However this does come at the cost of reduced sensitivity as the 8 and 10 mm diameter SAs have much larger resolution in pressure and resistance. To exploit the better sensitivity of the smaller diameter SAs, more SAs could be used in parallel to share the load and overcome the load-bearing limitations.

## **3.6 Biomedical Applications**

### **3.6.1 Probe**

The SA was first demonstrated in a probe application for identifying possible tissue abnormalities, helping to distinguish between objects and areas of different stiffness. Two different scenarios were investigated: in the first scenario, the SA was mounted on a rigid-bodied probe and used to distinguish between seven different objects of varying stiffness, used as representatives of the diverse tissues in the human body; and in the second scenario, the SA was shown on an actuated soft-bodied probe and used to press down on two different regions of jelly, one pure jelly and the other with a rigid object within, to mimic stiff nodules in tissue.



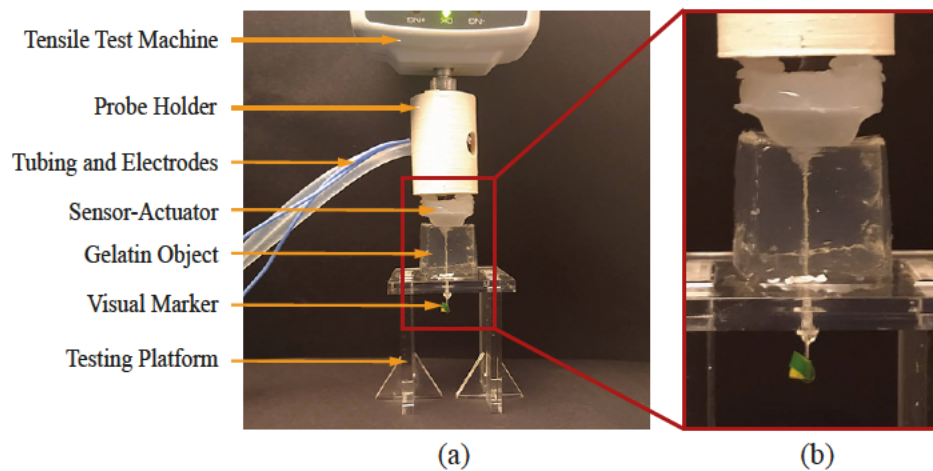


Fig. 3.10 The SA's integration in a rigid-bodied probe application. (a) Photo and (b) zoom-in of rigid probe application. The rigid probe was pressed on the 1:10 (gelatin:water) jelly object. The rigid probe was mounted on the tensile test machine for ground truth force measurements and a needle at the tip of the SA with a visual marker was used as a reference for displacement.

### Rigid-bodied Probe

For the rigid-bodied probe, the SA was mounted onto a 3D printed part (Prusa i3 Mk3), and attached to a tensile test machine (Imada MX2). The full probe was then pressed, at a speed of 0.01 m/min up to a maximum of 6 mm, onto seven different objects of varying stiffness, used as representations of the diverse tissues in the human body (Fig. 3.10). A solid acrylic plate was used as a baseline stiff object, and the other six objects were mixes of varying weight ratios of water and gelatin, with the softest being a mix of one part gelatin to 25 part water by weight, with a stiffness of 0.4 N/mm and the hardest with a mix of one part gelatin to two parts water by weight, with a stiffness of 13.7 N/mm. The predicted height was compared to a visual marker at the end of a needle fixed at the top of the SA and tracked using the Tracker software. During the experiment, the resistance and the pressure were also monitored using the developed system, and used to predict a force and the height of the SA. The prediction was made using the multiple linear regression model, derived above, combining pressure and resistance readings. The predicted force was compared to the force monitored using the tensile test machine, acting as a ground truth for validation. The 10 mm diameter SA with 1.5 mL of saline solution was chosen for the application as it was among the most consistent, with a good height and loading resolution, as well as a good inflated bubble height.

In Fig. 3.11, representative results from a subset of the probe experiments are shown, with the mean predicted and ground truth forces and SA heights plotted for the maximum

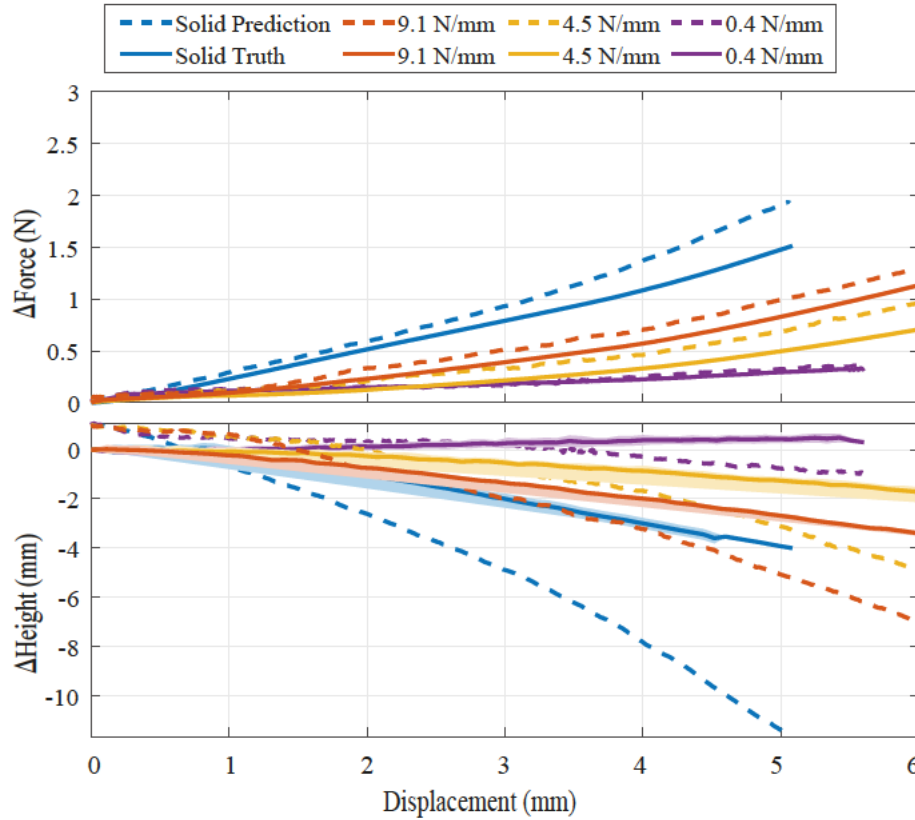


Fig. 3.11 The SA's validation in a rigid-bodied probe application. (Top) Predictions of force and ground truth measurements for the three objects tested for the displacements of the tensile test machine. (Bottom) Predictions and ground truth of difference in height of the SA for the displacements of the tensile test machine. The probe with the SA at the end is indented on a seven different objects, from solid objects to different ratio mixes of water and gelatin, with a subset of four representative objects shown here. The tensile test readings are used as a reference measurement for force and a needle at the end of the SA is used as a reference for height, with the model from the characterisation experiments used to predict the force and height from the resistance and pressure readings. The solid lines show the mean across five trials, with the shaded region representing one standard deviation.

6 mm displacement of the probe, and standard deviation shown by the shaded region. The subset included the solid object and gelatin objects mixed with ratios of gelatin to water of 1:5, 1:10 and 1:25, corresponding to stiffnesses of 9.1 N/mm, 4.5 N/mm and 0.4 N/mm respectively. Overall the probe, with the resistance and pressure measurements, shows a good ability to distinguish between the soft and stiff objects, even for objects of small stiffnesses, with a clear difference in slope similar to the ground truth. However, for the stiffer objects, the force predictions are greater than the measured forces, with an error of 0.5 N for the solid object. On the other hand, for the softer objects the gap between the predicted and measured

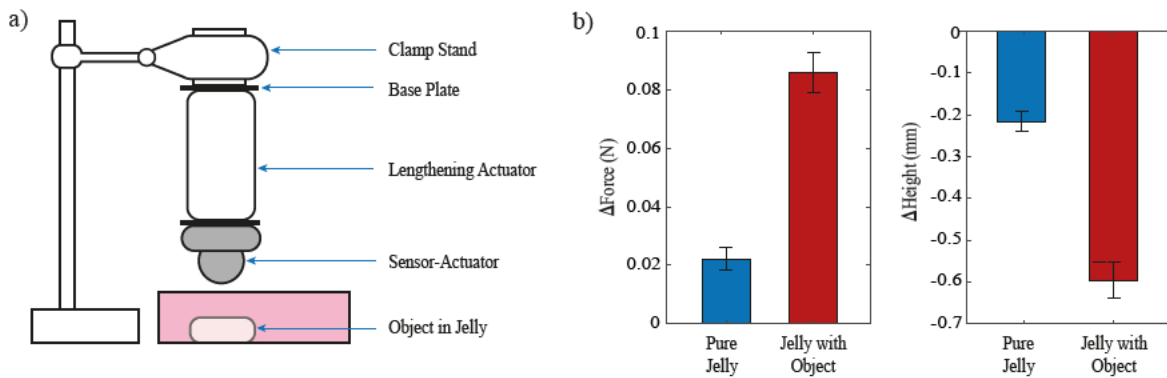


Fig. 3.12 The SA's integration in a soft-bodied probe application. (a) Schematic of the soft SA mounted on an axially lengthening soft implant actuator. The soft probe is pressed into the jelly using the axially lengthening soft actuator. (b) Results of the (Left) predicted force and (Right) height change for the SA, as the soft-bodied probe was pressed into a pure jelly region and a region of jelly with an embedded 3D printed object. The error bars show the standard deviation across five trials.

forces is small, less than 0.1 N. For the difference in height predictions, the SA over-predicts the difference by more than double and also has an initial offset of 0.5 N, likely from a drift in resistance between the modeling data and the experimental data, but does show a good relative distinction between all of the objects.

### Soft-bodied Probe

Following these tests, the probe application was further extended in a test on a fully soft-bodied probe, to distinguish two regions in a jelly substrate. For this experiment, the SA was mounted on an axially-lengthening soft actuator and pressed into a pure jelly (Hartley's) region and a region where a 3D-printed PLA object was embedded within the jelly (Fig. 3.12(a)). The linear soft actuator (detailed and developed further in Chapter 4) was used as a platform to generate a 3 mm extension for the SA validation and demonstrate easy integration with a soft robotic platform. Both the SA and the axially lengthening soft actuator were fixed in place with a base plate and held above the jelly object with a clamp stand. For the testing, the axially-lengthening actuator was inflated with saline solution to allow the inflated SA to press into the different jelly regions. The 10 mm diameter SA with 1.5 mL of saline solution was again chosen for the application, and both resistance and pressure were monitored.

The maximum difference in predicted force and predicted height show a clear distinction for the two different regions tested (Fig. 3.12(b)). The region with a 3D printed object, caused a much larger difference in both predicted force and height, nearly 0.1 N and  $-0.6$  mm,

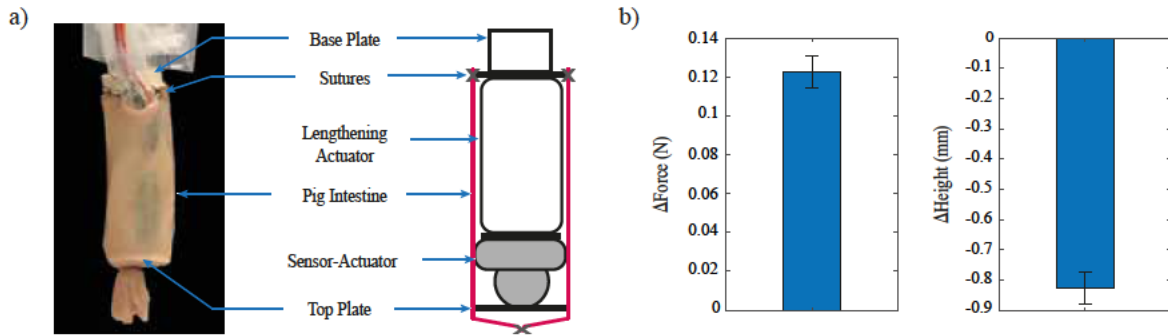


Fig. 3.13 The SA's integration and validation in a soft robotic implant. (a) Image and schematic of the assembled implant robot, the SA and the axially-lengthening soft actuator, in a pig intestine. The top end of the implant was fixed in place with a base plate and simple interrupted sutures. The other end was fixed by a top plate, with the tissue tied off below it with suture thread, to create a flat surface for the soft implant to push against the tissue. (b) (Left) Maximum predicted force and (Right) height change from the SA over time, as the implant was progressively lengthened and then retracted repeatedly using the axially-lengthening soft actuator. The error bars show the standard deviation across five trials.

compared to the pure jelly region which caused very little change, 0.02 N and  $-0.2$  mm. The soft-bodied probe is therefore clearly able to distinguish between the two different regions, although does suffer from slight drift, possibly from electrolysis or changes in temperature, throughout the experiment, which is reflected in the error bars.

### 3.6.2 Soft Robotic Implant

Finally, the SA was tested as part of a robotic implant, and used to monitor the length grown and the force applied to a pig intestine (Fig. 3.13(a)). For the experiment, the SA was mounted onto the same axially-lengthening soft actuator and in this application, the base plate was attached with simple interrupted sutures to the pig intestine. The other end of the implant, was fixed by a top plate, used as a flat interface with the tissue, and the intestine tissue was sealed off below the top plate with suture thread.

In Fig. 3.13(b), the mean and standard deviation of maximum force and height displacement results from five cycles of lengthening and retraction of the soft robotic implant are shown. As the whole soft-robotic implant is inflated, the SA at the top is progressively compressed, leading to a predicted maximum  $-0.8$  mm change in height and also progressively applies up to a predicted 0.12 N of force on the tissue. As the whole implant is soft, some of the force could also be being absorbed by the soft axially-lengthening actuator, as well as some tolerances in the integration and assembly of the two parts of the implant, so the



true force applied to the tissue could be greater, and would still need further experiments to confirm. Also, the measurements do again experience some slight drift, which is reflected in the standard deviations. Overall, the soft robotic implant, with the SA included, provides an indication of the force and height change experienced by the surrounding tissue. Further experiments are still needed to provide a clearer idea of the true forces and height change experienced by the soft robotic implant as a whole; the accuracy of these values is challenging to validate due to the soft robot - soft tissue interaction.

### 3.7 Conclusion and Future Work

Overall, this work presents a compact and versatile soft ballooning SA, a key building block for many different robots, that is capable of measuring displacement and force, essential to estimate tissue lengthening and stiffness, in both static and dynamic conditions. The investigative study explored the advantages, differences and complementarities of two sensing modalities, pressure and resistance, and found them to be interchangeable, although best used in combination. The static characterisation showed good repeatability across trials and demonstrated the interchangeability of using either the pressure or the resistance to predict height and load, although the predictions perform best when combining both readings. The load predictions also performed better overall, compared to the displacement predictions, which was also noticed in the application example. In future, more investigation is needed to improve the predictions, extending on the angled loading experiments, to understand more complex interactions, such as with uneven objects or when the SA is subject to multiple 3D forces, such as a stretching force in addition to the loading. This would be particularly relevant for stacked SAs that bend and elongate, where different SAs would be subject to uneven loading during bending [72]. Adding more electrodes and using electrical impedance tomography, could be another way to create a more complete understanding of the SA shape under more complex interactions.

For the dynamic characterisation, again there is good resolution between conditions up to 1.5 mL volume. For the larger volumes, the pressure and resistance readings show inconsistent behaviours, from lateral expansion and overall lengthening of the actuators. To improve the results, a stiffer base could be used, to constrain this undesired behaviour. Additionally, the membrane could also be changed to a stiffer material or alternatively be used in parallel with other SAs, if stiffer tissues are to be targeted in probe or gripper applications, or if larger forces need to be applied, such as for tissue lengthening applications, to prevent the flattening of the SA and achieve a similar shape to the current SA. The interface between

the tubing and the SA also requires more investigation, as it is the main failure point of the SA.

Finally, the probe application demonstrated a good ability to distinguish between objects of different stiffness. For the rigid-bodied probe, the predictions of force using the multiple linear regression models from the static characterisation were comparable to the tensile test machine results. The predictions of displacement were less good though, with some notable gaps between the prediction and ground truth, although the general trend is preserved. Additionally, for the soft-bodied probe, the SA was able to distinguish clearly between soft and stiff areas. Finally, for the soft robotic implant application, the SA was able to predict forces and displacements experienced by the SA, but further work is still needed for better integration and to distinguish between the force experienced by the SA compared to the actuator as a whole. This distinction could potentially be achieved by adding similar sensing modalities of the SA to the axially lengthening soft actuator, but needs investigating. Overall, using volumes within the 1.5 mL range, and benefiting from the better resolution, would be reasonable for the applications, however drifts in resistance of up to  $1\text{k}\Omega$  between experiments days apart and less than  $0.1\text{k}\Omega$  during around 30mins of testing, would need to be accounted for. This is also one of the main factors impacting performance and importantly predictability, so better calibration and more controlled testing is needed.

Additional research into more thermally stable biocompatible fluids, such as liquid metals and or potassium iodide glycerol solutions [197], needs to be conducted, and could be used to reduce drift problems. Alternatively, a temperature sensor could be added to the system and more data collected at different test temperatures, to help provide better training data for the estimations. The dynamic ability of the SA could also be explored to enable re-calibration during long procedures, by deflating the SA and resetting the starting resistance, minimising issues from drift. The ability to change the liquid throughout the experiment, with changes in volume and/or concentration, could also allow the dynamic tuning of the sensor properties and aid in characterizing the SA itself and its surrounding environment, by allowing multiple measurements with an SA with different sensing properties to be taken and improving the predictions. The replenishable nature of the SA could also counter any effects of electrolysis of the ionic solution, as well as mitigate any evaporation of the solution due to the permeability of the silicone. The fabrication of the SA equally needs attention, with investigation into alternative manufacturing techniques needed to further improve the SA's manufacturing consistency, as well as allow further miniaturisation of the design. The use of less permeable silicones could also be considered to minimise leakage, especially if the SA is to be left inside the human body for long periods of time.

Finally, to integrate the SA into existing surgical technologies a number of challenges need to be overcome. For use in an implant, the size of the control electronics should be made portable, which could be achieved using medical grade syringe drivers or a smaller custom design, additionally maximizing axial lengthening, to allow maximal tissue lengthening is crucial, so different ballooning shapes could be considered. For a probe application, interfacing with existing probe systems would need to be considered. Initial efforts on interfacing with 3D printed and soft probes have been shown in this work, but a sturdier attachment would need to be considered, potentially using progressively stiffer soft materials to smooth the transition. Scale and portability of SA and surrounding platform would also need to be considered, again using a mixture of custom and off-the-shelf components. In the case of sensing for grippers, the channeling of the fluid and cables on the gripper claws, without compromising or hindering movement would be essential, as well as attachment to the ends. For all applications, additional resistive sensing electrodes could be added to improve the resolution in the more uncontrolled environments, at the cost of additional fabrication complexity. To finish, further testing in more realistic scenarios is needed, with the addition of bio-fluids, as well as further tests on biological tissues for the probe applications.

## 3.8 Chapter Summary

Achieving compact and biocompatible advanced technologies with sensing capabilities is a key challenge for the safety critical and highly patient-specific biomedical field. In this chapter, a compact and versatile soft fluidic sensor-actuator (SA) capable of measuring both force and displacement in static and dynamic conditions is presented. Pressure and resistance are shown to be interchangeable, although best used in combination, when predicting the load on the SA, and show good repeatability and distinction between the loaded and constrained conditions. Using a single sensing medium and across the different diameters tested, the best estimated resolution of just under 4 g and 0.07 mm is achieved with the 12 mm sensor, filled with 1.5 mL using pressure sensing only. Furthermore, the SA is demonstrated in two probe applications and as part of a soft robotic implant for tissue-loading based tissue regeneration. The SA showed the ability to distinguish between different objects or areas of varying stiffness, as part of both a rigid-bodied and soft-bodied probe, as well as being able to predict force and displacement from the lengthening and retraction of a soft implant. Overall, this SA has the potential to be a key building block for biomedical robots' monitoring of both displacement and force.





# Chapter 4

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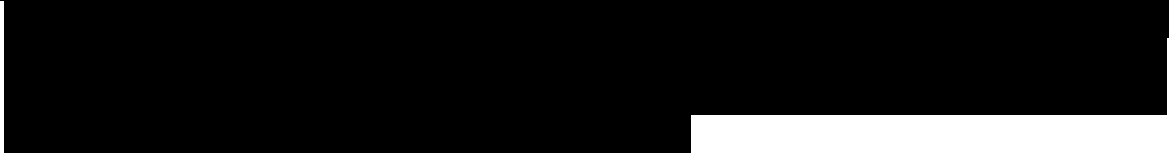
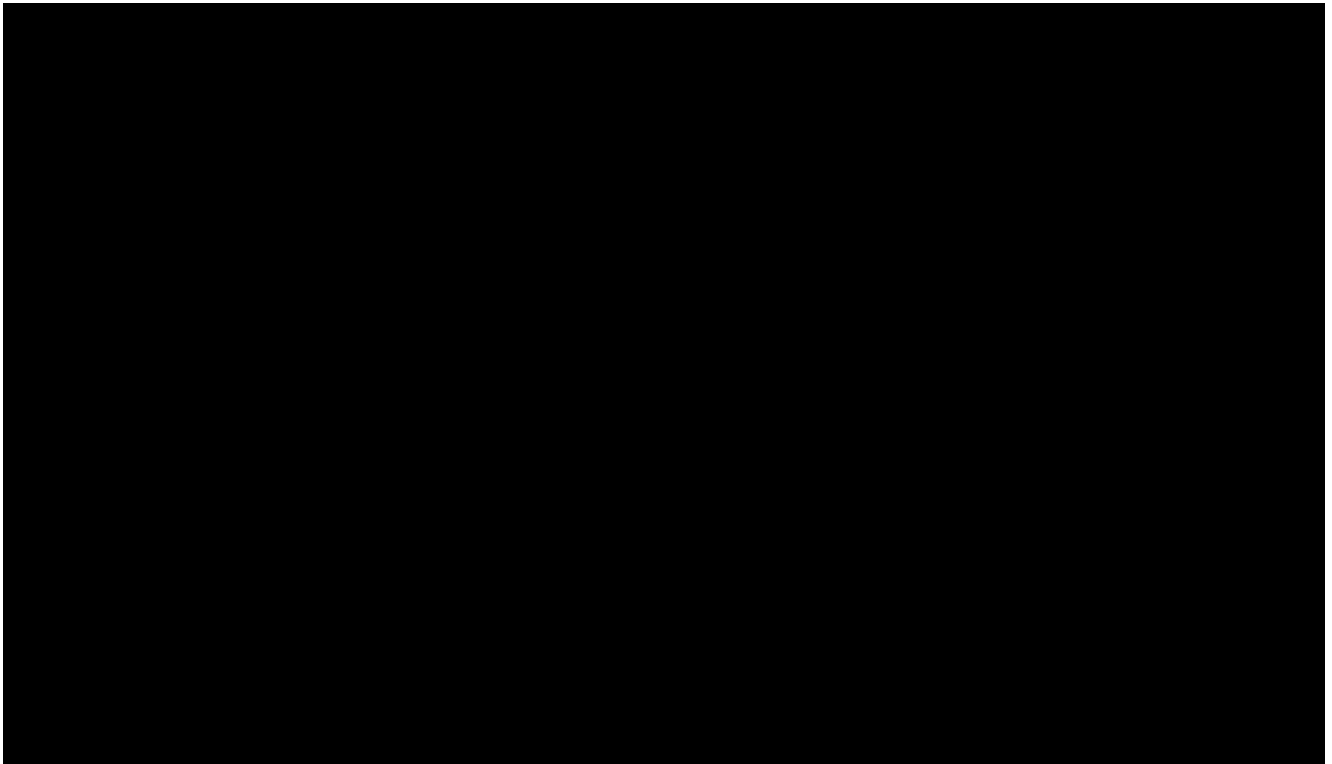
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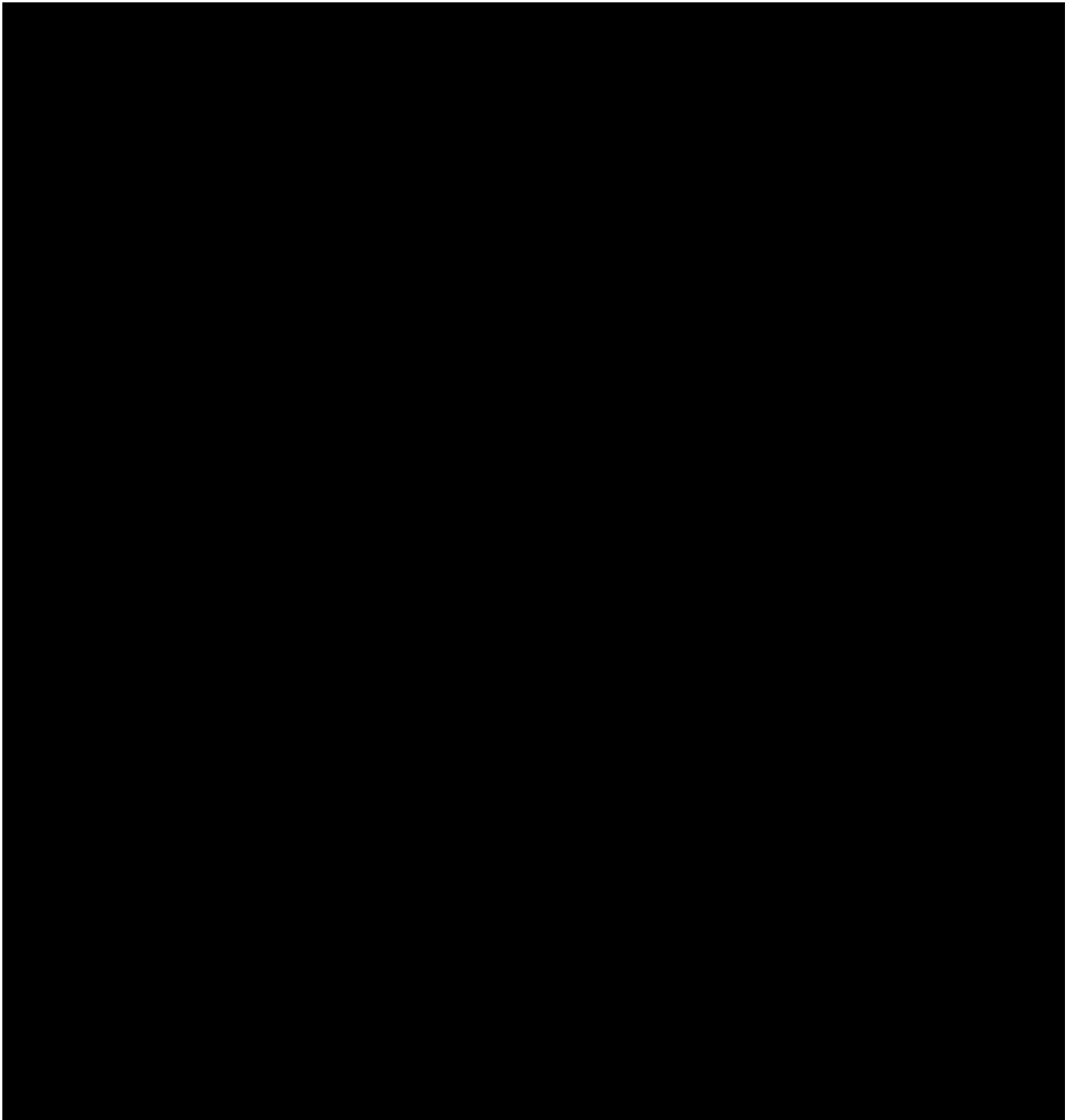


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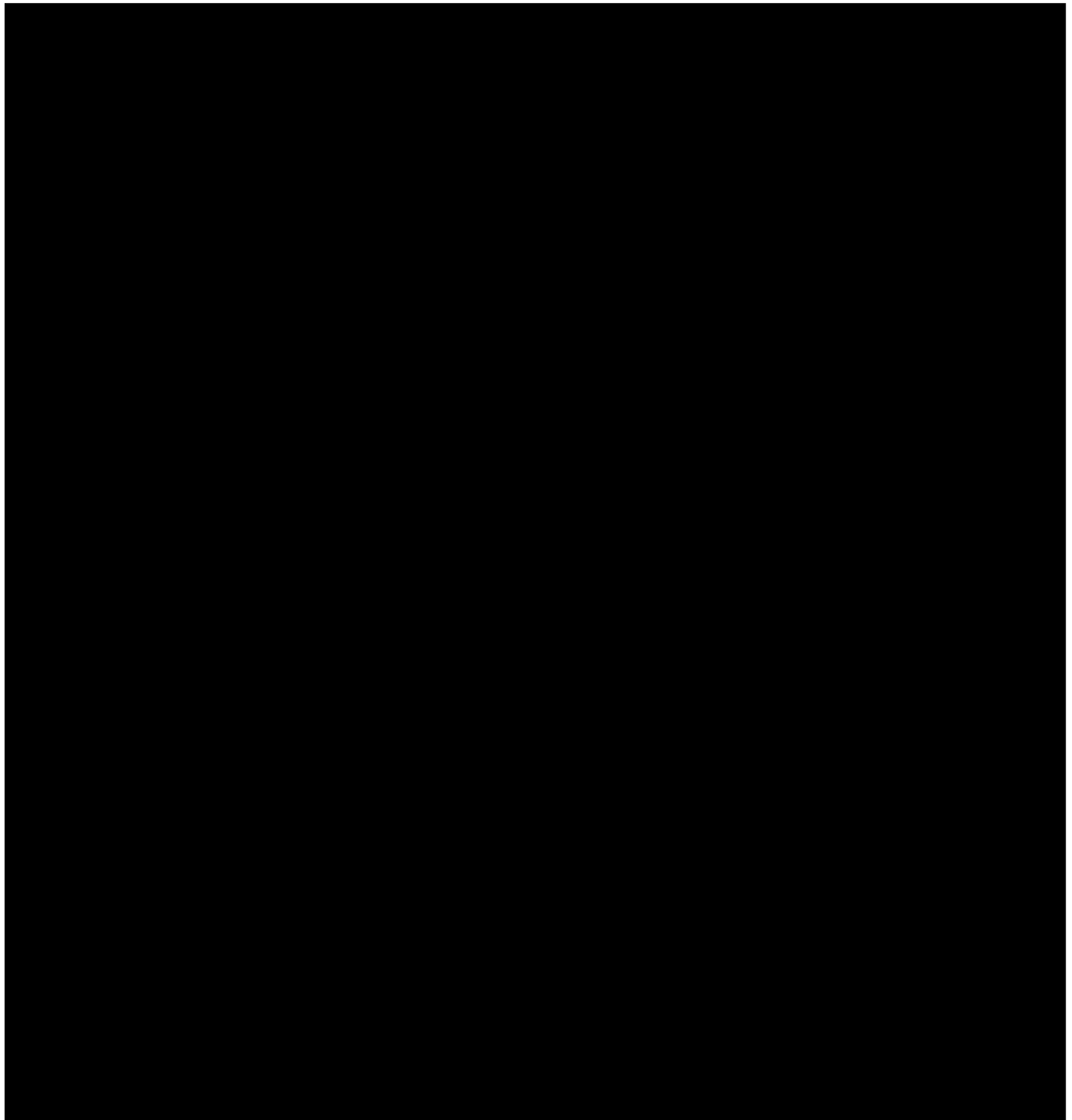
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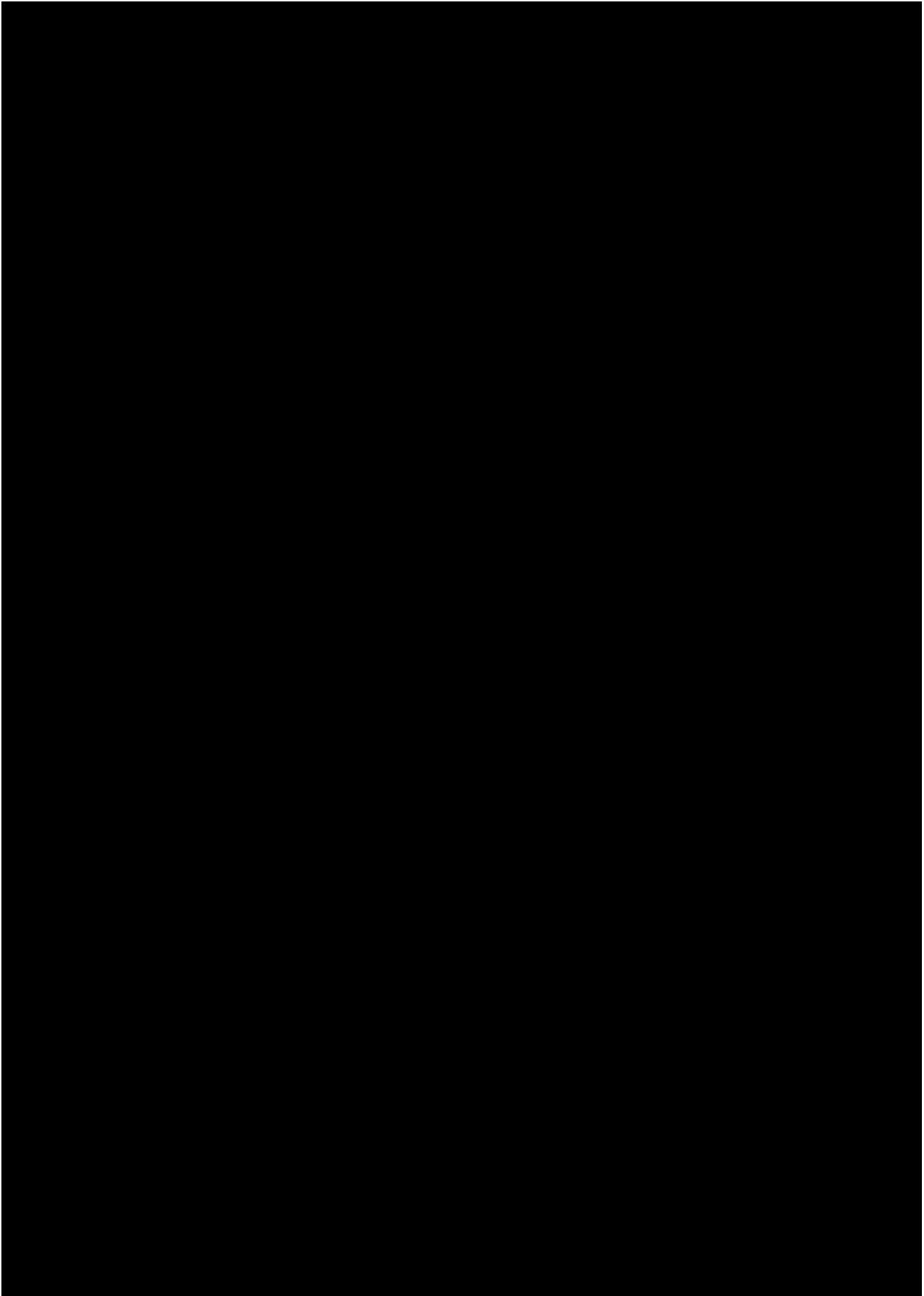
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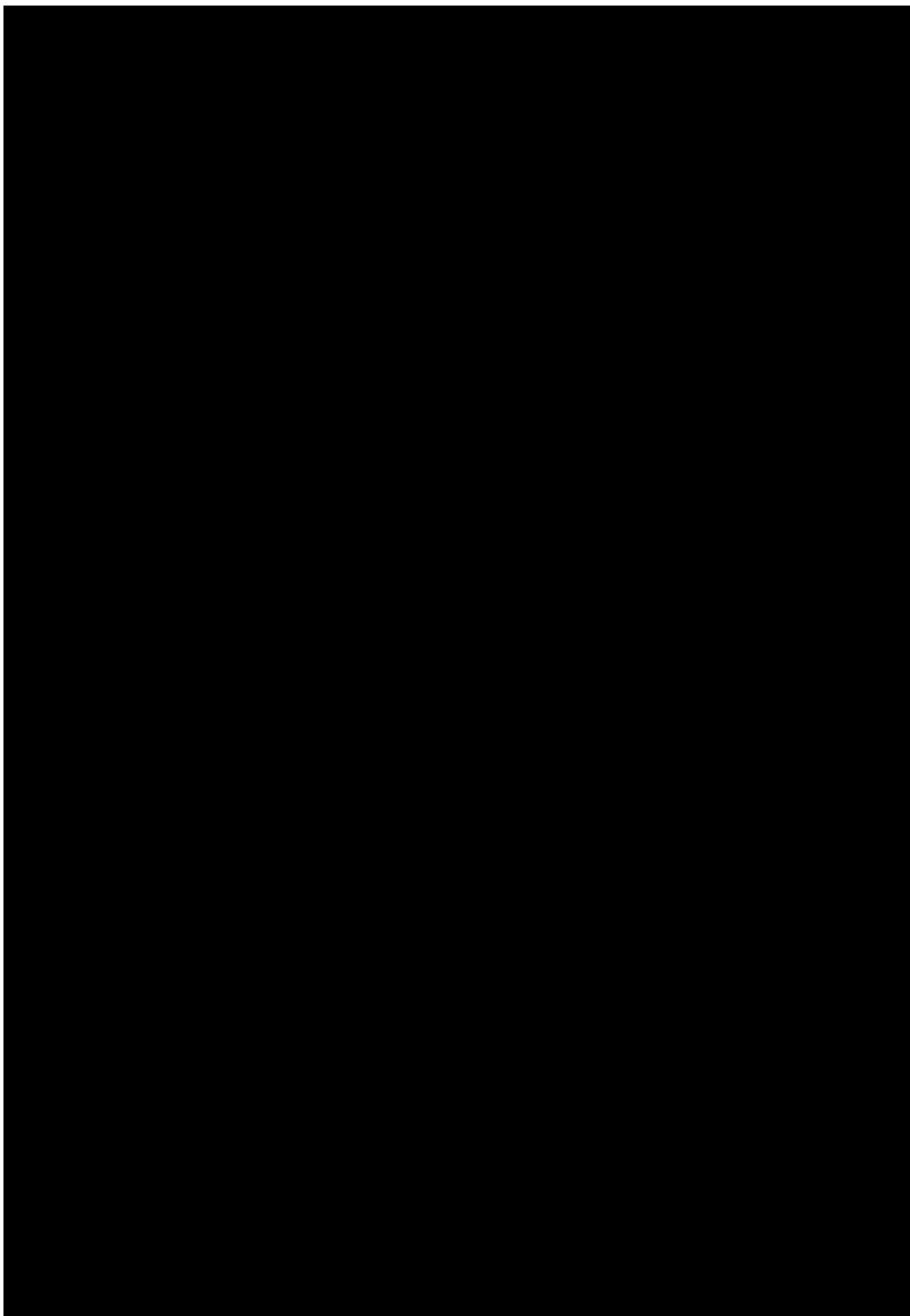
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# **Chapter 5**

## **Tuning of and Localisation in Soft Strain and Pressure Sensors**

### **5.1 Preface**

In this final technical chapter, the work from [83] and [84] is presented sequentially. The use of ionic solutions in sensors is explored, taking advantage of their biocompatibility and suitability for in-the-body medical application, as well as exploiting the dynamic potential of fluid as a sensing medium.

In the first work, a soft strain sensor that can be embedded into soft actuators, such as shown in Fig. 5.1[138] is presented. Its resistance characterisation for two different volumes of ionic solution and two different concentrations is shown, demonstrating the potential to tailor the sensor depending on the range of motion to be detected. This was a preliminary study and further testing in applications was not carried out. In the second work, a pulsating fluidic sensor (PFS) capable of distinguishing and differentiating between multiple press events of various forces, sizes and locations is presented. Made up of just two parallel fluid channels and one pair of sensing electrodes, the PFSs present a compact and versatile solution for information-rich sensors. The soft PFS is also demonstrated in two sensing applications with relevance to rehabilitation and assistive wearables.

The contributions of the chapter include: (1) the concept, design, and fabrication of soft sensors; (2) basic modelling of the soft strain and pressure sensors; (3) the characterisation of a soft strain sensor, whose electrical properties are tuned, by varying its volume and its ionic concentration; (4) the characterisation and testing of localisation in a soft pressure sensor, exploiting the dynamic changes induced by a fluidic pulse; (5) the demonstration of a soft

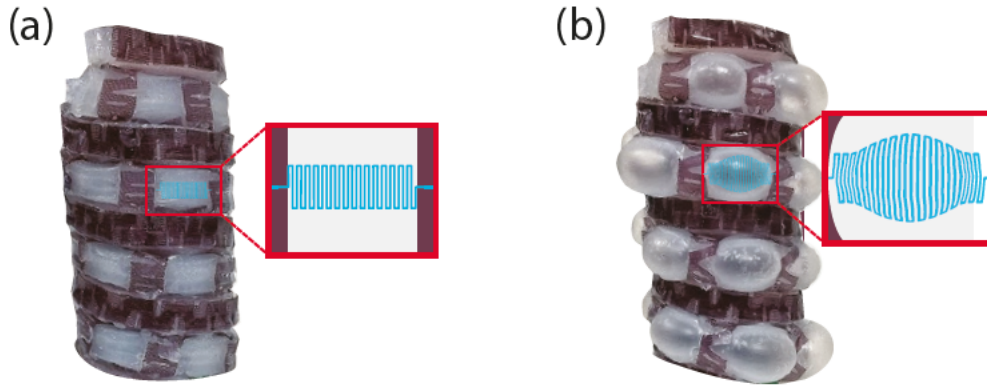


Fig. 5.1 Conceptual integration of the soft sensor in an implantable soft sensor for tubular tissue regeneration to sense strain/pressure on the tissue [138].

pressure sensor, able to distinguish the location and size of different presses in two wearable applications.

The first work was published as part of [83] and reproduced with permission from Springer Nature. The second work is ©2024 IEEE. Reprinted, with permission, from [84].

## 5.2 Introduction of a Soft Strain Sensor

Soft robots are being increasingly used in the medical field, with applications in surgery, wearables and implants [32, 140, 138]. With their inherent compliance and potential biocompatibility, they are ideal for interacting with the human body. Additionally, given the demands of healthcare technologies and the required accuracy, there is a clear need for soft sensors to increase the autonomy of soft robots without compromising their mechanical properties [100]. Extensive work on soft sensors capable of proprioception and/or pressure sensing based on synthesized conductive elastomers has been carried out [190]. There have also been recent developments in the use of ionic solutions and hydrogels, as equivalent and biocompatible conductive liquid for the sensors [78, 71], thus getting closer to soft robots being implemented within the human body. The use of ionic solutions for sensing has been previously explored by investigating piezoresistive effect throughout elastomeric models. The effect of varying parameters such as the input AC frequency or concentration has been studied previously [71], but little work has been done on varying the volume within the same sensor.

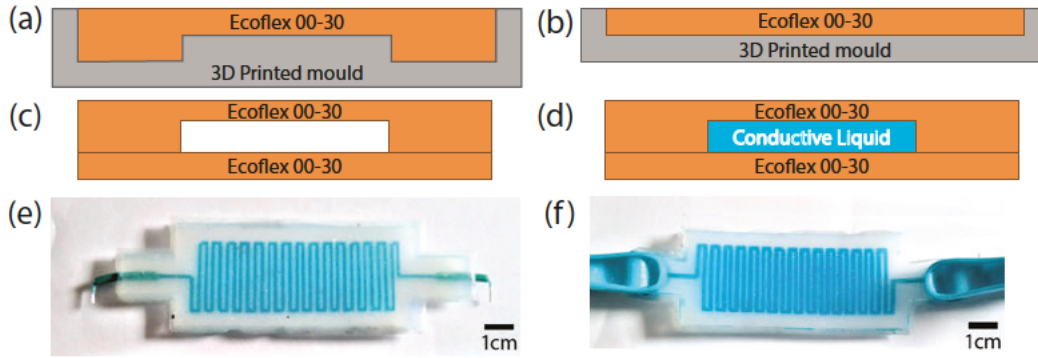


Fig. 5.2 Schematic of the major fabrication steps. (a) and (b) the layer with the channel and the top layer respectively are moulded and cured separately. (c) The layers are joined using uncured Ecoflex 00-30. (d) The ionic solution is injected. (e) and (f) Photo of the assembled sensor with 1mL and 3mL of ionic solution respectively.

## 5.3 Materials and Methods

### 5.3.1 Fabrication

The soft sensor used in this first study was inspired by the commonly used serpentine strain sensor design, optimising for maximal lengthening and resistance change in a small footprint. The 1 mm channel dimension was chosen for manufacturing ease, and the overall sensor was designed to be 10 cm long. 3D printed moulds printed on a Form2 3D printer were used to cast the soft sensor layers. These layers were made out of Ecoflex 00-30 (Smooth On Inc.), mixed using an ARE-250 Mixer (Thinky), degassed and then poured into the moulds curing at air temperature (Fig. 5.2). The Ecoflex was then post-cured in the oven for two hours at 80°C and one hour at 100°C. Finally, the sensor layers were bonded together using uncured Ecoflex 00-30, after which the ionic solution was injected and sealed using copper leads. The general dimensions of the sensor are 100x30x3mm, with a 1mm<sup>2</sup> cross-sectional area serpentine channel with an active area of 700mm<sup>2</sup>.

### 5.3.2 Experimental Procedure

The sensor was tested under tension using a Zwick Roell Z020 tensile test machine at a speed of 20 mm/min. (Fig. 5.3). For the experiments, two different concentrations - 25% and 33% of NaCl to water by mass - and two different volumes - 1mL and 3mL - were tested. A 15kΩ resistor was connected in series with the soft sensor, and monitored using a NI USB-6009 DAQ board at a frequency of 2.4kHz. The circuit was connected to a Gwinstek SFG-1013 function generator and supplied with a 1V 1kHz voltage sine wave.

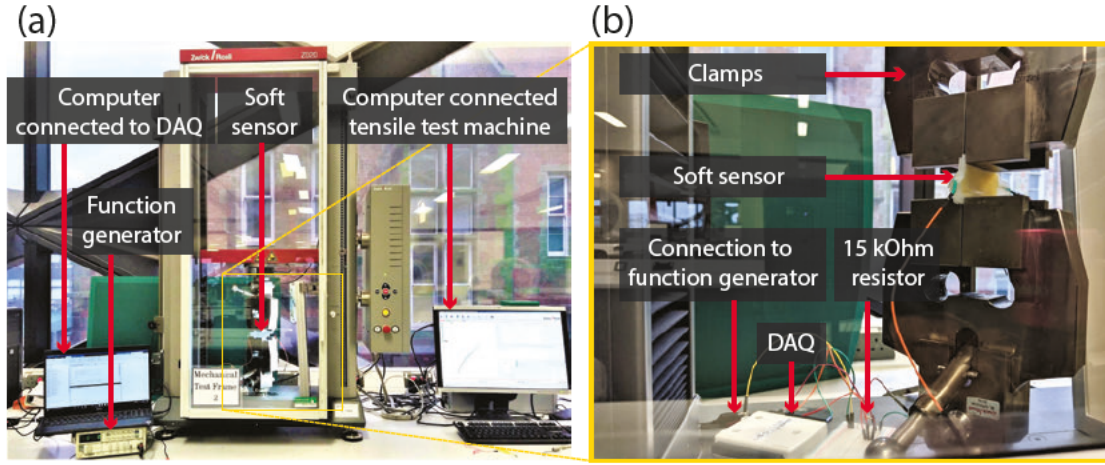


Fig. 5.3 Experimental Setup (a) Full setup. (b) Enlarged view of the soft sensor attached to the tensile test machine.

The rectified voltage signals were filtered using a moving average with a window size of 240 points. The sensor resistance was then calculated using the voltage divider equation:

$$R_{\text{sensor}} = R_{15k\Omega} \times \frac{V_{\text{supply}} - V_{\text{mes}}}{V_{\text{mes}}} \quad (5.1)$$

Where  $R_{\text{sensor}}$  is the resistance across the sensor in  $\Omega$ ,  $R_{15k\Omega}$  is the resistance of the sensor in series,  $V_{\text{supply}}$  is the supplied voltage from the function generator and  $V_{\text{mes}}$  is the voltage measured by the DAQ.

## 5.4 Modelling

In addition to getting experimental results, a model was also derived for the resistance during stretch. The basic law describing changes of resistance depending on the material and the geometry is Pouillet's law:

$$R = \rho \frac{l}{A} \quad (5.2)$$

Where  $R$  is the resistance in  $\Omega$ ,  $\rho$  is the resistivity in  $\Omega m$ ,  $l$  is the length in  $m$ ,  $A$  is the cross-sectional area in  $m^2$  and  $V$  is the volume in  $m^3$ . For ionic solutions of 25% and 33% concentration the conductivities are  $0.0450\Omega m$  and  $0.0414\Omega m$  respectively.

For the strain sensor, assuming a constant volume  $V$ , Eq. (5.2) can be rewritten as:

$$R = \rho \frac{l^2}{V} \quad (5.3)$$

Where  $l$  changes as the sensor is stretched vertically.



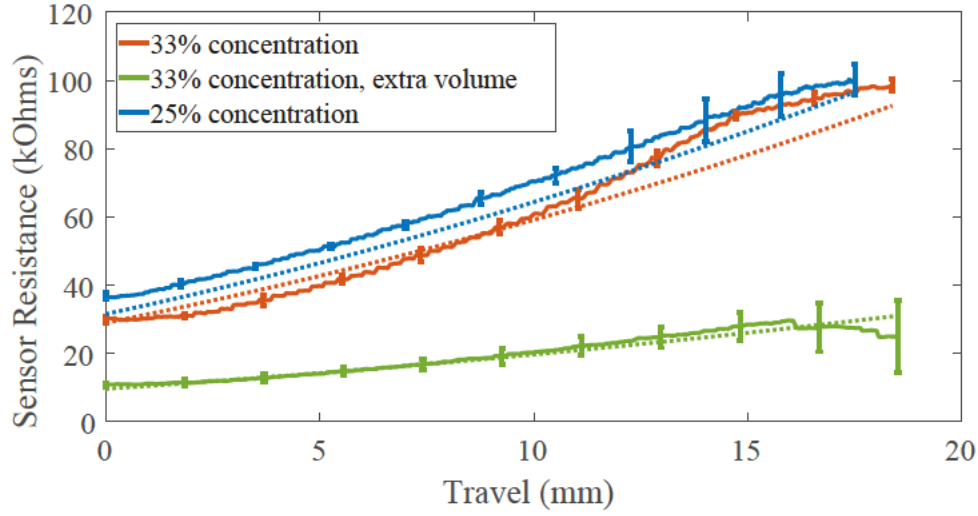


Fig. 5.4 Sensor resistance vs displacement for the three different testing conditions. The error bars represent one standard deviation across three trials.

The new length of the channel  $l$  after stretch can be described by:

$$l = N * (l_0 + \Delta l) + l_{\text{unstretched}} \quad (5.4)$$

Where  $l$  is the new channel length,  $N$  is the number of vertical channel segments,  $l_0$  is the original vertical (direction of stretch) segment length,  $\Delta l$  is the travel distance/stretch distance and  $l_{\text{unstretched}}$  is the length of the horizontal segments (not in the direction of stretch), assumed to be fixed in length.

## 5.5 Results

The experimental and modelling results are shown in Fig. 5.4. There is a good agreement between the model and the experimental values, with the general trends and order of magnitudes being similar. Both the experimental and modelling results show the average increase in resistance as the sensor was stretched to 20mm for the baseline concentration of 33% and volume of 1mL, as well as the increased volume of 3mL and the decreased concentration of 25%. The sensor with decreased concentration has a resistance of roughly 10kΩ higher than the baseline sensor throughout the travel. The resistance for the sensor with increased volume is around a third of the resistance as the baseline sensor for the same travel.

## 5.6 Discussion

A decrease in the sensor resistance with increased volume is observed, as well as an increase in resistance when the concentration decreased. The decrease in resistivity for the increased volume is in accordance with Pouillet's law and the model, which states resistance as inversely proportional to the cross-sectional area, although more work needs to be done to characterise this effect fully. The large variation of resistance for an increased volume is also promising. With the resistance ideally equal to the series resistor, the sensor with increased volume would be more accurate for larger displacements, while the sensor with less volume would be more accurate for smaller displacements. Overall, the results also highlight the ability to vary the measured resistance greatly with a change in volume, compared to the small change for an increase in concentration.

## 5.7 Conclusion and Future Work

We present and characterise a soft sensor to be integrated as the skin of soft implantable robots. The results in the study are promising and show a potential to tailor not only the concentration, but also the volume of ionic solution for the desired sensing range. As future work, we will investigate techniques to vary the liquid composition of the soft sensors to automatically tune its characteristics. This is desirable when the environment in which the soft robots operate changes. A simpler sensor design, of a tubed sensor with ionic solution, could also be tested, with the same modelling principles being applicable.

## 5.8 Introduction of a Soft Pressure Sensor

Information-rich and space-efficient sensors are key requirements for many applications, biomedical ones in particular. In the medical context, understanding the environment robots are placed in is essential for safe patient-robot interaction [32]. Receiving information on constraints in the operational space and on how the environment reacts to interactions represents a critical requirement for robots to make informed control decisions. For assistive and rehabilitation devices, improving portability, by minimizing the resources needed for sensing, as well as achieving compact designs, to not hinder the natural movement of the body, are crucial. These have proven to be stringent requirements challenging current sensor technology development [190].

Accurate and precise sensing in soft robotics is commonly accomplished using multiple sensors. However, achieving their seamless integration into robots and wearable devices,

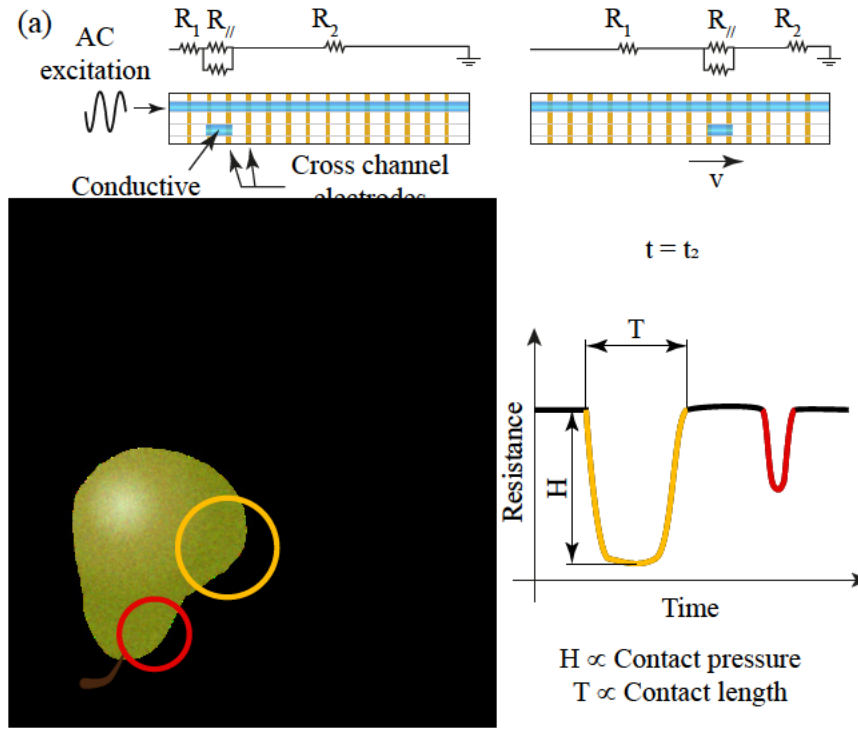


Fig. 5.5 Conceptual image of the pulsating fluidic sensor (PFS) and its sensing capabilities. (a) Schematic representation of the PFS and its electrical equivalent. (b) Demonstration of the PFS on a robotic hand to sense location, pressure and contact area of a gripped object and the resulting temporal change in resistance.

without compromising human or robot motion, adding excess weight for users, or being computationally demanding, still requires further exploration. Most common soft sensors have been dedicated to transducing pressure and stretch. This has been achieved using resistive, capacitive, piezoresistive, optical [192] and magnetic [152] technologies, mostly embedded in fluidic or conductive stretchable substrates. Distinguishing between stimuli is also a key area of research and, in fluidic sensors, tends to rely on multiple layers of resistive-based conductive liquid layers, with sensitivity to various stimuli derived from the channel shape, capable of differentiating between pressure and strain events [95][62], or between normal and shear forces [187]. These sensors are more compact and require less external hardware, but provide only general information and usually lack localisation.

Traditionally object localisation detection has been approached using sensor arrays to capture the spatial distribution of pressures, but these often require multiple active electrodes, fast processing speeds, and are not easily integrated into systems [165][104][128][111]. More recent approaches have relied on optimizing the placement of sensors and creating regions of targeted sensing, using a combination of software [178] and prior knowledge of the robot's application and potential movements [37][30][168]. These techniques have been

implemented with the aim of limiting the number of sensors needed, and general bulkiness of the sensing setup. However, these strategies rely on knowing movements and behaviours beforehand, limiting their applicability across multiple users or for humans interacting with dynamic environments, where behaviours and reactions are unpredictable. For soft robots, work has also been done towards combining sensing and actuation to further limit the size of sensing elements [71][81], but again, the localisation requires bulky external setups using multiple electrodes [144].

More recent work focusing on achieving localisation within a minimal footprint has been carried out, using a variety of different modalities. Acoustic waveguides and optical ultrasound have been studied to investigate the detection of a variety of different stimuli, including strain and deformation, and have shown potential in localizing contacts, but require fast processing speeds [28][31]. Similarly, research using two pressure sensors at the ends of a closed air medium to locate fault events from signal timing has been conducted, but both the high frequency of data sampling needed and the difficulty detecting small non-damaging press forces present limitations [60]. Distributed magnetic sensing has also been used to transduce continuous deformation produced by stimuli, with relevance to localisation and shape reconstruction, but requires rigid external circuitry to be placed in close proximity to ensure operation [70][199]. Using fluidic sensing, some work has been conducted towards localisation, using two parallel conductive channels with opposing sensitivities, and differentiating the varying stimuli from a comparison across the channels, but it is neither capable of detecting multiple stimuli events, nor characterizing the size and force of the press events [93]. Work using a single channel, with additional band-pass filter components embedded in the soft sensors, has also been demonstrated to distinguish location of events, but its compliance and scalability is compromised with the necessary additional rigid electronics [94]. There is therefore room for a scalable fluidic sensor with minimal hardware for localisation of single and multiple press events.

## 5.9 Materials and Methods

### 5.9.1 Design concept of the soft PFS

The working principle of the soft PFS, shown in Fig. 5.5a, relies on two fluidic channels electrically connected with equidistant wires, with the spacing determining the sensing resolution, but limited by manufacturing capabilities. One channel is fully filled with a conductive ionic liquid and excited using a sinusoidal electrical signal, while the other enables the laminar flow of a conductive fluidic pulse (segment of ionic solution trapped

between air gaps). With the laminar flow being crucial to ensure the fluidic pulse does not break down and hinder the resistance measured, but also restricting the potential sensing speed and the sensor's inner diameter. During a press event, the cross-sectional area of both channels is reduced. When the pulse passes through this area, the difference in resistance is amplified. The location and size the press events cause direct changes in the timing, and duration of resistance spikes respectively. In addition, the larger the force being applied on the sensor, the greater the amplitude of the spikes. Furthermore, for multiple pressure points, the number of resistance spikes increases, and the location of these spikes is reflected in their timing.

The sensor can be modelled as a simple resistive network as shown in Fig. 5.5a. Thus, this pulsating fluidic conductive circuit allows the spatio-temporal identification of external stimuli deforming the sensor, by inducing resistance changes during its travel along the channel (Fig. 5.5b).

### 5.9.2 Fabrication of PFS

The fabrication begins by fixing two lengths of silicone tubing (ID:1.5 mm and OD:2.5 mm) into a sewing jig, designed to aid with aligning the elements of the sensor. A length of thin copper wire is then sewn across both silicone tubes in a zigzag pattern spaced 2.5 mm apart. After the excess loops of wire are cut to separate the electrodes, the connected tubes are carefully removed from the jig and the ends of each electrode are folded over the tubing. Both tubes with the connecting electrodes are subsequently fixed in a resin 3D-printed mould (Form 3, Formlabs) and Ecoflex 00-50 silicone is poured over the top for final sealing and fixing. Before being poured, the Ecoflex 00-50 is first mixed in a mechanical mixer (ARE-249 Mixer, Thinky), and then degassed for three minutes. Once poured, the Ecoflex 00-50 is left to cure at room temperature (approx. 20°C) for at least three hours. Finally, the sensor is removed from the mould, the tubing ends are trimmed, and the sensor is filled with an ionic solution (9 g salt for 1 L distilled water). The first channel is completely filled with the ionic solution and plugged with electrodes on both ends. The second channel is left empty and is connected to a fluidic setup and two timing electrodes that enable the tracking of the fluidic pulse throughout the experiments. The fabrication steps are shown in Fig. 5.6.

### 5.9.3 Characterisation Setup and Experimental Protocols

The full experimental setup used is shown in Fig. 5.7. The experimental setup consists of the PFS under testing, a signal generator and a DAQ connected to sensing and timing electrodes, as well as a custom-built pressure-driven flow control system for the fluidic pulse. A tensile

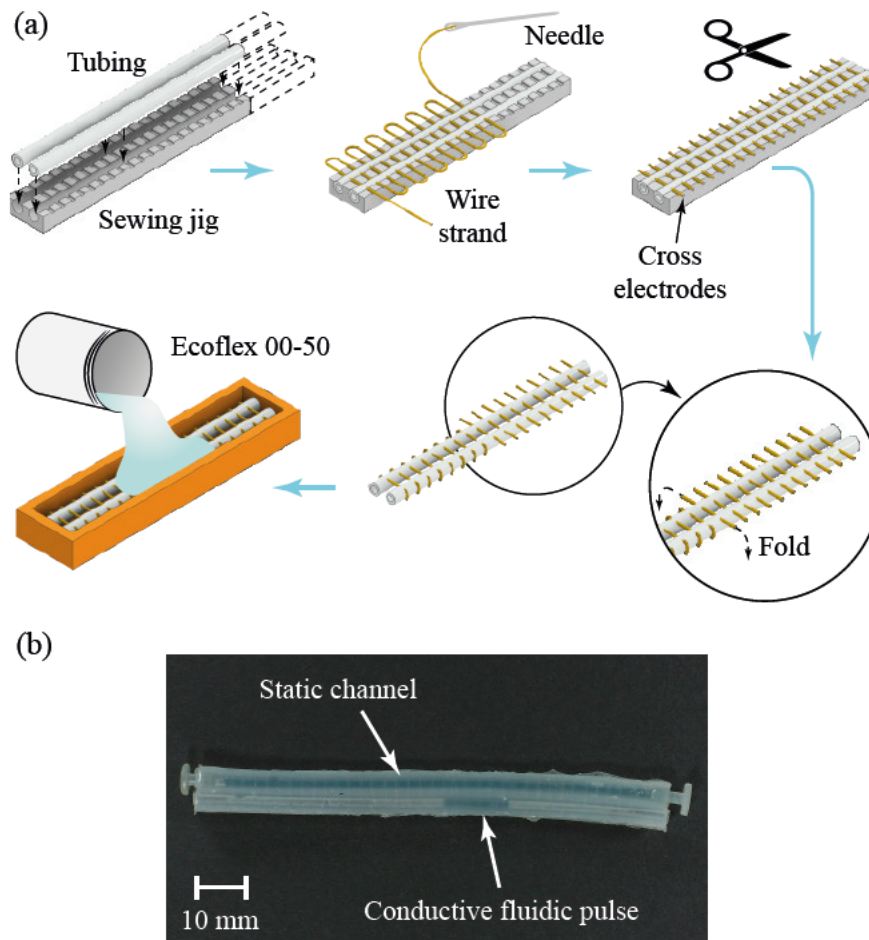


Fig. 5.6 Fabrication. (a) Schematic of the fabrication steps of the pulsating fluidic sensor (PFS) and (b) the final photo of the fabricated and filled PFS. The fabrication involves: fixing two lengths of silicone tubing into a custom sewing jig; evenly sewing wire across both tubings; trimming the edges of the wires to separate into individual electrodes; folding the electrodes over the tubings; and sealing and fixing the tubings with Ecoflex 00-50.

test machine (Imada) or calibrated weights were used to carry out the press events and monitor the force amplitude of the press, as well as carry out different lengths and numbers of presses using a variety of custom attachments. During the press events the sensor itself was connected to multiple electrodes, all made by stripping 5 mm of the plastic covering and soldering the end of a 24 AWG copper wire, to monitor changes in resistance and to track the fluidic pulse. The fully filled channel was plugged by two electrodes, supplied with a 1 V, 1 kHz sine wave and monitored using a custom-built DAQ acquiring the signal at 20 kHz. The fluidic pulse was also tracked by two extra pairs of electrodes at both the inlet and outlet, connected to the same signal generator and DAQ. The purpose of these was to calibrate the timing of the saline solution segment, but are surplus to requirements. Finally, throughout the



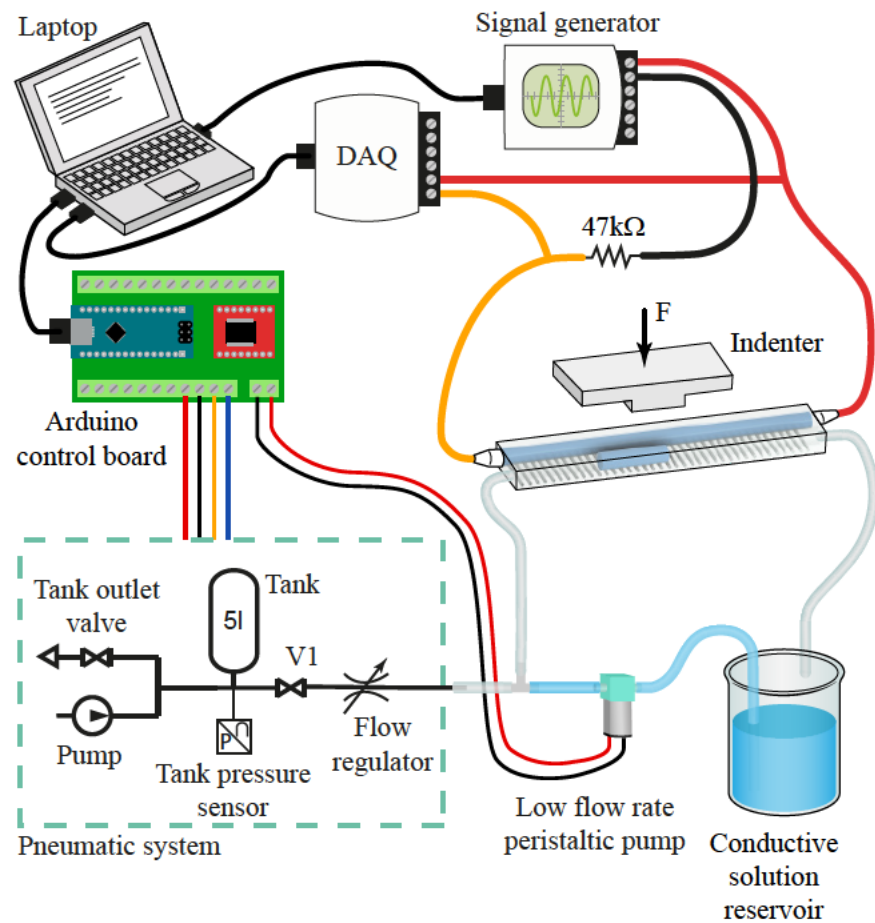


Fig. 5.7 Experimental setup. The experimental setup consists of the sensor connected fluidically to a peristaltic pump generating the fluidic pulses along with a pneumatic system driving the flow of the pulse, and electrically connected to a voltage divider circuit supplied by a signal generator and monitored with a DAQ. Throughout the experiments the sensor is pressed with an indenter, with varying forces, indent sizes and across multiple locations.

experiments, a pneumatic setup was used to push the fluid segment generated by a peristaltic pump (Dolomite) at an average speed of 2.3 cm/s. This was achieved by pressurizing a 5 L reservoir tank to 60 kPa and reducing the pressure with a flow regulator as shown in Fig. 5.7.

The PFSs were tested under multiple press conditions, including different locations, sizes and force amplitudes of press. Five different press forces of 1 N to 5 N in 1 N increments were examined (same press size), as well as four different press sizes of 0.5, 1, 2 and 3 cm, each pressed down with 50 kPa. Five different press locations, from 1 cm to 9 cm evenly spaced by 2 cm, across the sensor (press force of 5 N) and multiple simultaneous press events were also tested on the PFS. These different conditions were chosen to show off the sensing and location capabilities of the PFSs.

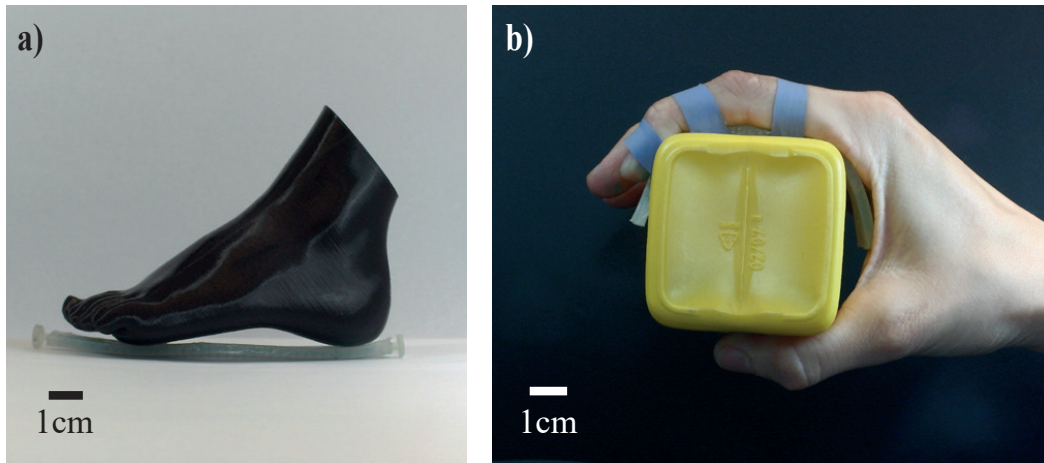


Fig. 5.8 Photos of application setups (a) Probing foot stance abnormalities, where A 3D printed foot is placed on the PFS. (b) Probing gripping with a PFS-embedded finger wearable. For the foot placement detection application, the foot was weighted evenly, on the toe and the heel and the location of the pressure was determined. For the grip location detection application, a cuboid object was gripped at varying angles and the location of the interaction of the finger with the object was determined.

#### 5.9.4 Applications

After its characterisation, the sensor was tested in two wearable applications, chosen over an implantable application for ease of implementation given the current size restrictions around the fabrication. The two applications were a foot placement detection and a grip location detection, with potential to determine stance and grip abnormalities respectively, applicable in rehabilitation scenarios. For the foot placement detection, a single sensor was tested underneath a 3D printed foot model and manually weighted. An even foot weighting, a toe and a heel foot weighting were tested and compared with an unloaded sensor. For the grip location detection, a single sensor was fixed to the underside of a human index finger, and the sensor was tested on both a straight and curved finger, as well as when gripping a cuboid object between the fingertip and the thumb, the proximal phalanx and the thumb and finally between the distal and proximal phalanges and the thumb.

### 5.10 Modelling

The sensor can be modelled with a simplified electrical equivalent consisting of a series of resistors with one resistor in parallel, with each resistor governed by Pouillet's law.

The following assumptions for the model were made the length of the sensor is fixed; the liquid displaced by the press events is evenly distributed across the rest of the channel and



assumed to be negligible; change in cross-sectional area of the channel from a press event is uniform across the width of the press. Throughout the travel of the pulse, when the fluidic pulse is not under the indent, the total resistance can be described by:

$$R_{tot} = (L_s - L_i - \alpha_f L_f) \frac{\rho_s}{A_c} + L_i \frac{\rho_s}{A_i} + \alpha_f L_f \frac{\rho_s}{2A_c} \quad (5.5)$$

where  $L_s$  is the length of the full sensor,  $L_i$  the length of the fluidic indent,  $L_f$  the length of the pulsating fluidic segment,  $\alpha_f$  the proportional amount of the pulsating fluid in the system (varying from 0 to 1 as the pulse enters and reversely as it exits the system),  $\rho_s$  the conductivity of the saline solution,  $A_c$  the normal cross-sectional area of a single channel and  $A_i$  the cross-sectional area of a pressed channel.

When the pulse is under the indent, the total resistance can be summarised as:

$$R_{tot} = (L_s - L_i - (1 - \alpha_{fi})L_f) \frac{\rho_s}{A_c} + (1 - \alpha_{fi})L_f \frac{\rho_s}{2A_c} + \alpha_i L_i \frac{\rho_s}{A_i} + \alpha_{fi} L_f \frac{\rho_s}{2A_i} \quad (5.6)$$

where  $\alpha_{fi}$  is the proportion of the pulsating fluid under the indent and  $\alpha_i$  is the proportion of the indent without the pulsating fluid.

Using the empirically-derived values of conductivity and the change in cross-sectional area, a simulation of a few examples of the PFS behaviour is shown in Fig. 5.9(b).

## 5.11 Results and Discussion

### 5.11.1 Detection of Press Sizes

A variety of different press sizes were tested, and the results are shown in Fig. 5.10. Overall, for each size of press event there is a negative spike in resistance compared to the baseline with no press event. The duration of the spike increases as the press size increases, as the pulse experiences a change in geometry for longer with the wider presses. Differences in press sizes of 1 cm or more can be differentiated, with clear distinctions between the 1, 2 and 3 cm press sizes. The variation in the amplitudes of the signals, despite the sensor bearing the same pressure, could be due to manufacturing errors or more uneven loading across the

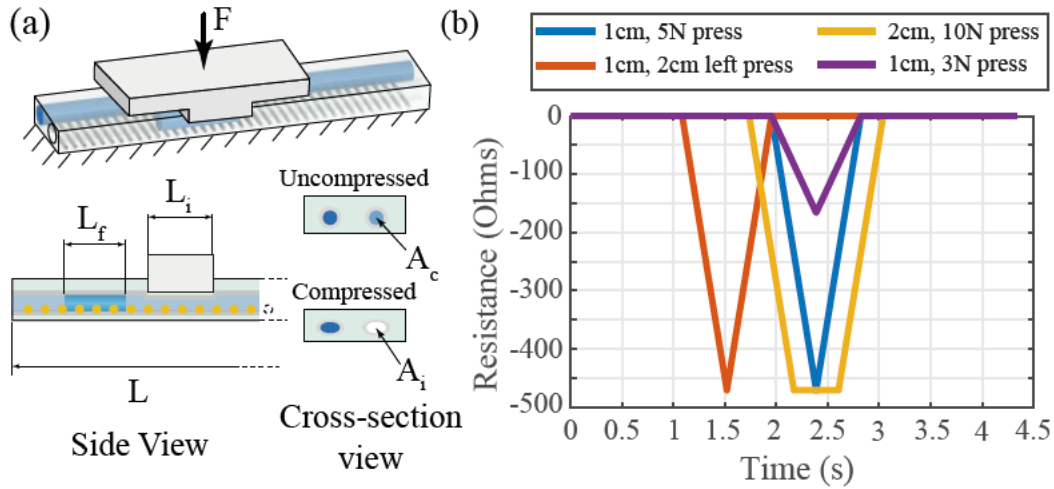


Fig. 5.9 Model. (a) Schematic of the sensor undergoing indent and key parameters. (b) Modelling results for different locations, sizes and forces of indents. Applying forces 2 cm further left results in a shifting of the time signal, applying wider indents results in a resistance spike longer in duration and applying smaller forces reduces the amplitude of the spike.

smaller surfaces. To conclude, the sensor shows a reasonable ability in distinguishing the size of press events.

### 5.11.2 Detection of Press Forces

The results from a variety of press force amplitudes from 0 to 5 N, on a 1 cm wide indent are shown in Fig. 5.11. The results show an increasing maximum drop in resistance as the force increases, visible across all press forces. An ANOVA test was carried out, which yielded that the minimum force that can be precisely identified among those tested is 2N, and similarly, the resolution of the PFS' force detection is 2N for forces under 3N, with the resolution increasing to 1N for forces higher than 3N. The resolution may prove larger if more granular force values were tested. The increased resolution for the larger forces could be explained by the tubing becoming more pronouncedly deformed beyond a threshold force. These large noise spikes result from the design with the cross-sectional electrodes being unevenly periodically connected as the pulse travels through the channel. Increasing the number of cross-electrodes would reduce these noise spikes and increase the distinction between the force amplitudes. Overall, the sensor demonstrates an ability to distinguish between different press forces.

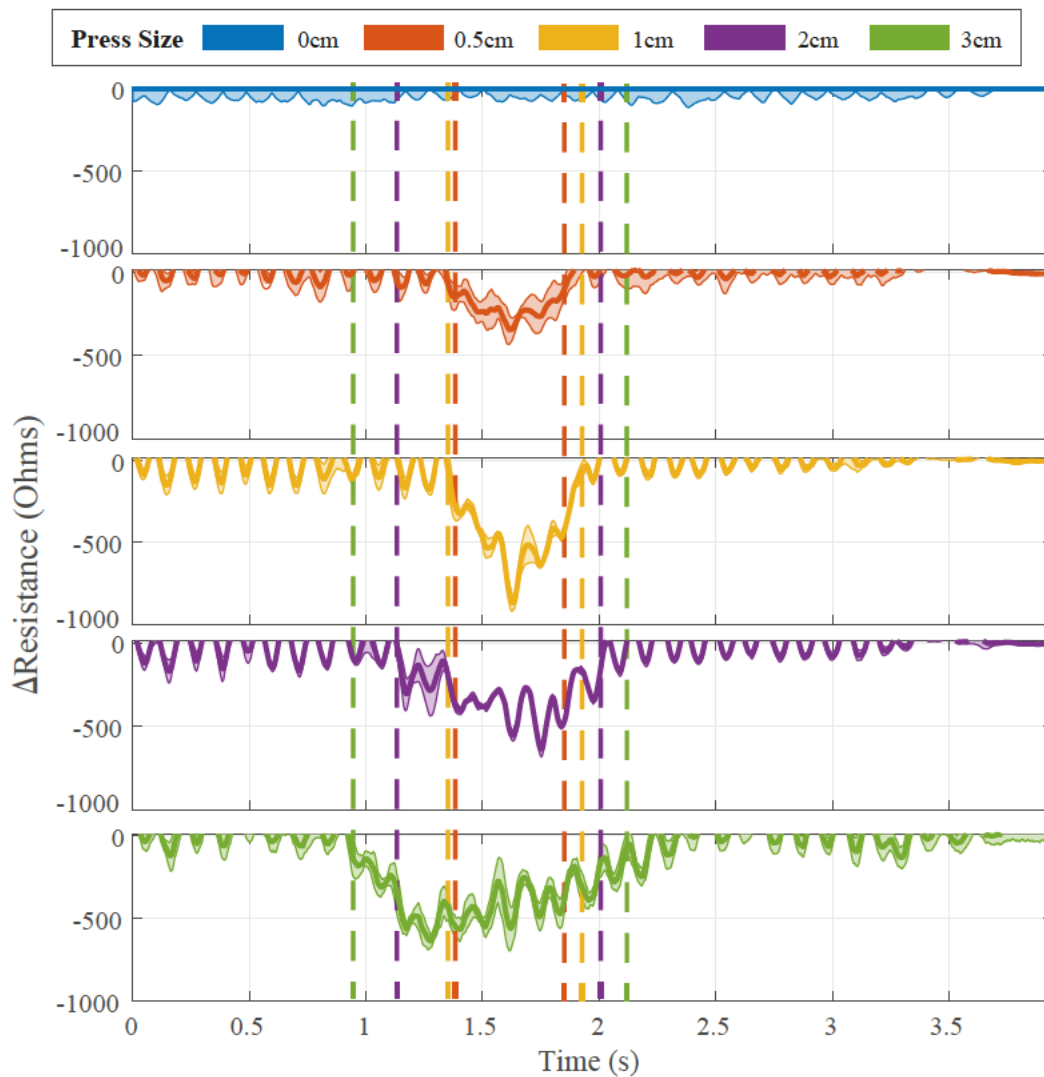


Fig. 5.10 Detection of press sizes. Various press sizes from 0 to 3 cm were tested. The average difference in resistance compared to the 0 cm press size and the standard deviation across five trials are shown.

### 5.11.3 Detection of Press Location

The location of a 5 N press with a 1 cm indent was tested across the length of the PFS and the results are shown in Fig. 5.12. The results show a good distinction between the different locations, with clear, and evenly spaced spikes across time, mirroring the evenly spaced indents tested. All spikes are similar in height, as would be expected, with the middle spike being the most pronounced at  $-250\Omega$ . The results demonstrate an ability to localize various press events from the timing of the spikes in resistance.

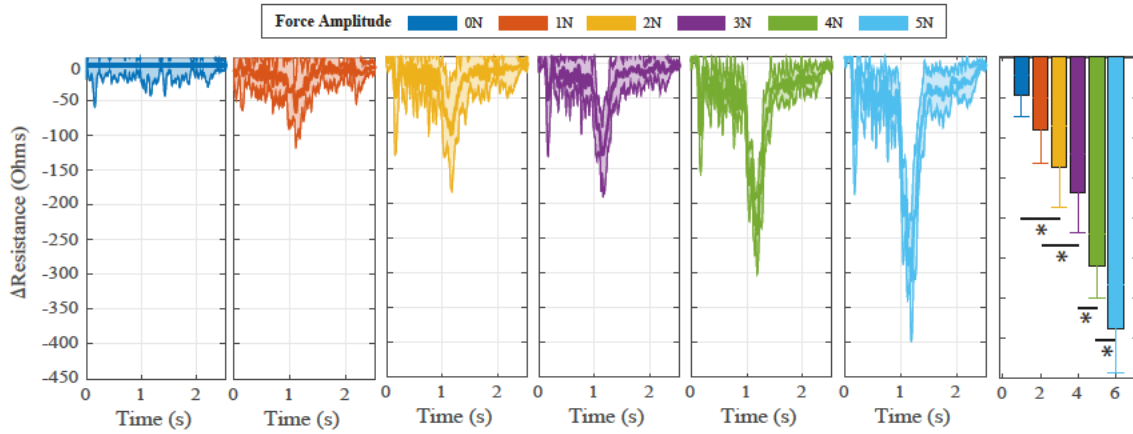


Fig. 5.11 Detection of press forces. Various press force amplitudes from 0 to 5 N were tested. The average difference in resistance compared to the 0 N press force and the standard deviation across five trials are shown. The asterisks in the bar chart highlight the closest neighbouring conditions where the mean difference in resistance amplitude is statistically significant.

#### 5.11.4 Detection of Multiple Presses

Building on the demonstration of using timing to identify location of indents, multiple indents were tested across the sensor and the results are shown in Fig. 5.13. An increasing number of press events from none to three were tested, each 1 cm in size, spaced evenly by 2 cm and loaded evenly with 50 kPa. From the results, subtracting the baseline change in resistance, the number of press events is clearly identifiable by the number of spikes in resistance. Additionally, the height of the spikes is similar across regardless of the number and similar for each of the conditions, with slight variations in peak resistance height from the experiments with two and three presses, likely due to uneven loading and distribution of the force across the indents. In general, the PFS demonstrated an ability to distinguish and locate multiple press events, exploiting the temporal changes experienced by the sensor as the fluid pulse travels along the sensor and amplifies notable changes in the geometry of the sensor.

## 5.12 Applications

Finally, the results from the two applications are shown in Fig. 5.14 and Fig. 5.15. For the foot placement detection, the results show peaks at the different foot locations that were weighted. For the evenly weighted foot, the results show two resistance spikes, with a larger spike of  $-1500\Omega$  at the heel and a spike of  $-500\Omega$  at the toe. With the toe weighted foot,

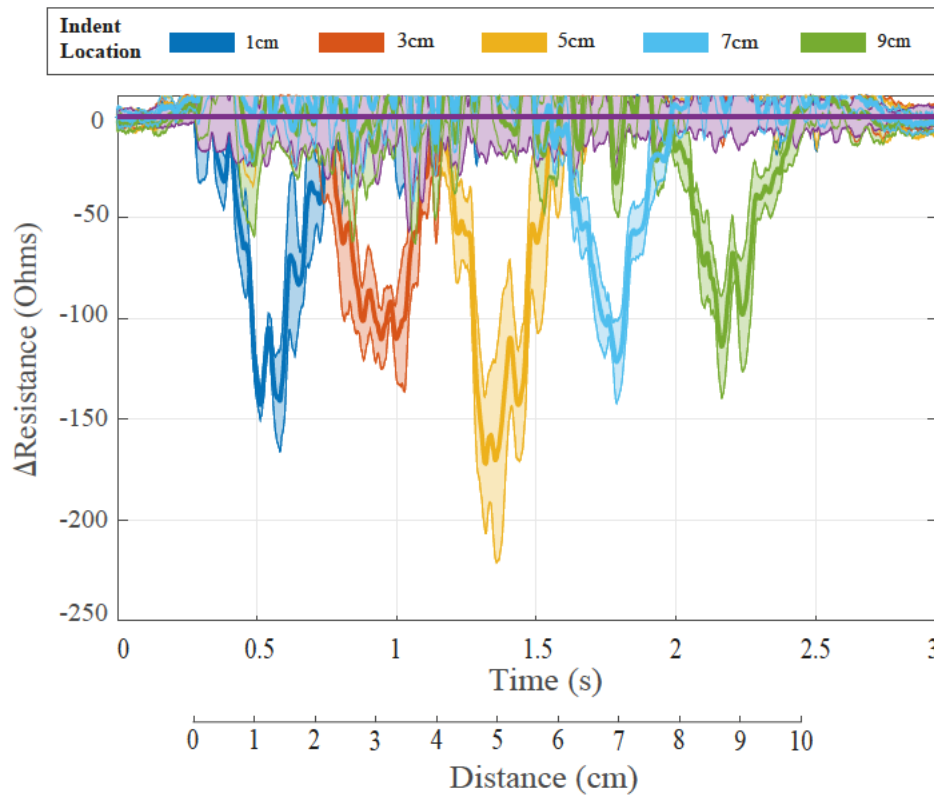


Fig. 5.12 Detection of press location. Various press locations, from 1 cm to 9 cm evenly spaced by 2 cm across the PFS, were tested. The average difference in resistance compared to a no press average and the standard deviation across five trials are shown. Distance estimates are also plotted accounting for the distance between the timing electrodes and the actual start of the sensor.

the spike occurs at the earlier location, with a similar amplitude of around  $-500\Omega$ . For the heel weighted foot, the spikes occurs at the later location, but with a much decreased amplitude of around  $-300\Omega$ . The smaller spikes for the uneven weighted feet could be explained from the difficulty of manually weighting at an angle, as larger spikes compared to the even weighted foot would have been expected.

For the grip location detection, the results show varying numbers of peaks and locations of peaks, matching the number and different locations where the finger and the object touch. In the application, the conductive fluid pulse travels from the palm of the hand to the fingertip. The spikes in resistance across the different grips are around  $-500\Omega$  and are sufficiently distinctive. Furthermore, curving the finger seemed to have little effect on the resistance, although stronger grips and curves might have a larger effect, with the more acute angles risking kinking and generally affecting the geometry of the channels.



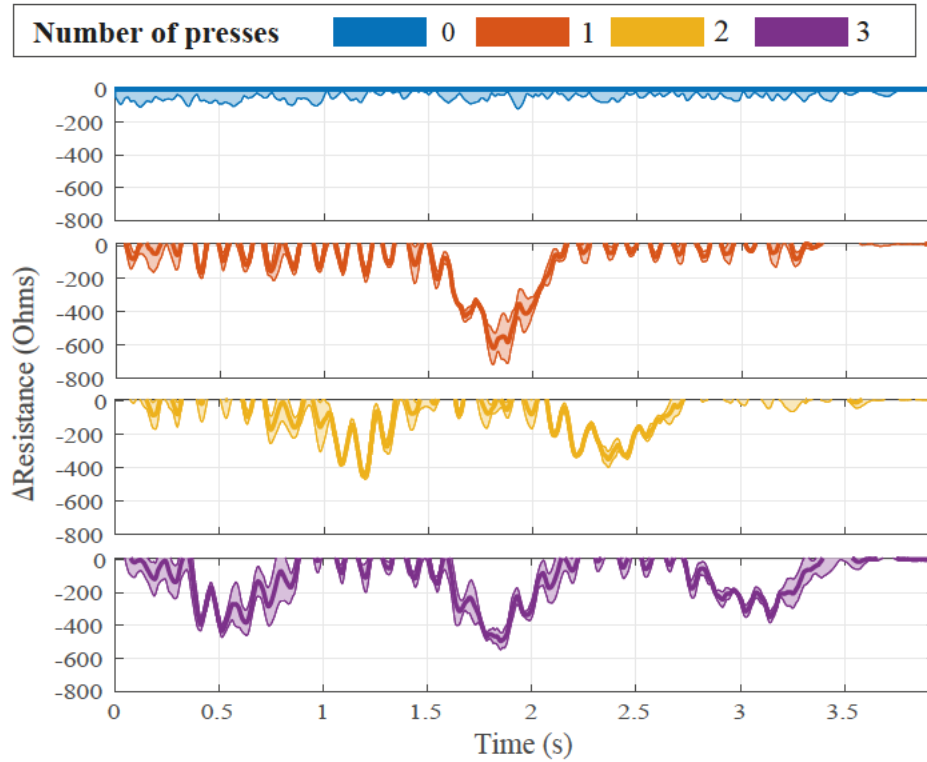


Fig. 5.13 Detection of multiple presses. One, two and three presses, each of 1 cm and spaced 2 cm apart, were tested. The average difference in resistance compared to a no press average and the standard deviation across five trials are shown.

### 5.13 Conclusion and Future Work

To conclude, this work presents a compact yet information-rich soft sensor, which can be used to detect the number, the size, the force and the location of press events. The characterisation highlighted the ability to localize single and multiple press events, by exploiting the spatio-temporal excitation of the system, as well as the potential to differentiate various sizes and force amplitudes of the press events. These characteristics were fully exploited and explored in the two applications presented, where the sensor was shown to be able to localize the weighting of a 3D printed foot, and distinguish between different grips, from the location of press event detection.

The overall sensor and setup does present some limitations. Notably the speed of the sensing, requiring on average 3 s to monitor the full 10 cm length of the sensor. Future work could include miniaturizing the channels further to achieve higher flow rates while maintaining laminar flow, therefore preserving the integrity of the conductive pulse. The use of extra conductive segments, with multiple running through the sensor, could also be explored to help increase the detection speed. Careful consideration of the size, distance

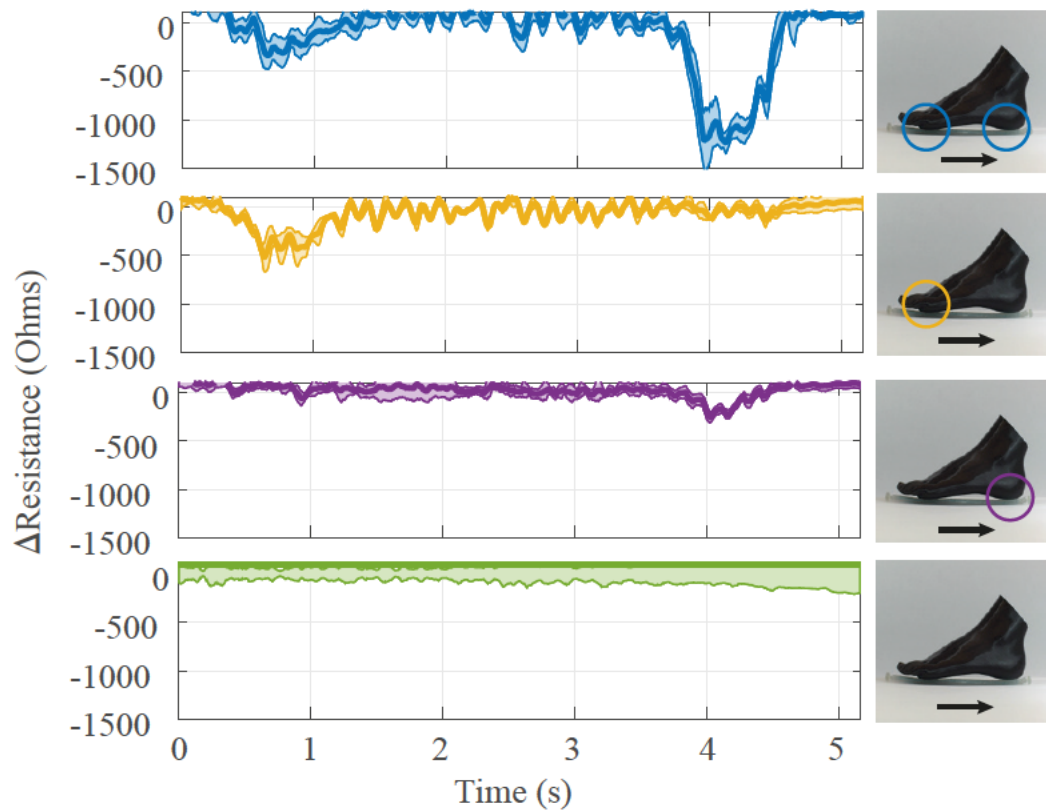


Fig. 5.14 Foot placement detection application. A 3D foot was weighted evenly, on the toe and on the heel. The average and standard deviation across three trials, relative to the baseline of an unloaded sensor, are shown.

and conductivity of the pulses would need to be taken though, to not lose localisation information. The precision and accuracy of the system overall could be improved by using a laminated approach and disregarding the heavily manual steps. Increasing the number of cross-electrodes or ultimately replacing the fully filled channel with a conductive elastic material, would improve both the signal-to-noise ratio and further minimize the overall size of the sensor. In addition to the sensor being miniaturized, the overall experimental setup also needs to be miniaturized further, using off-the-shelf fluidic pulse generators, to expand the potential applications to a greater number of robots. Recycling the pulse back through the sensor could also help with sizing down, although the manufacturing will need to be improved to not lose any liquid from being caught by the electrodes. Finally, a next major step is the development of automatic detection framework, using thresholding for example, to enable the prediction of the nature of the indentation from resistance changes. For the extension to more complex cases, such as overlapping events or more dynamic events, it might be possible to develop a reversible model using the deconvolution of the output resistance measurement

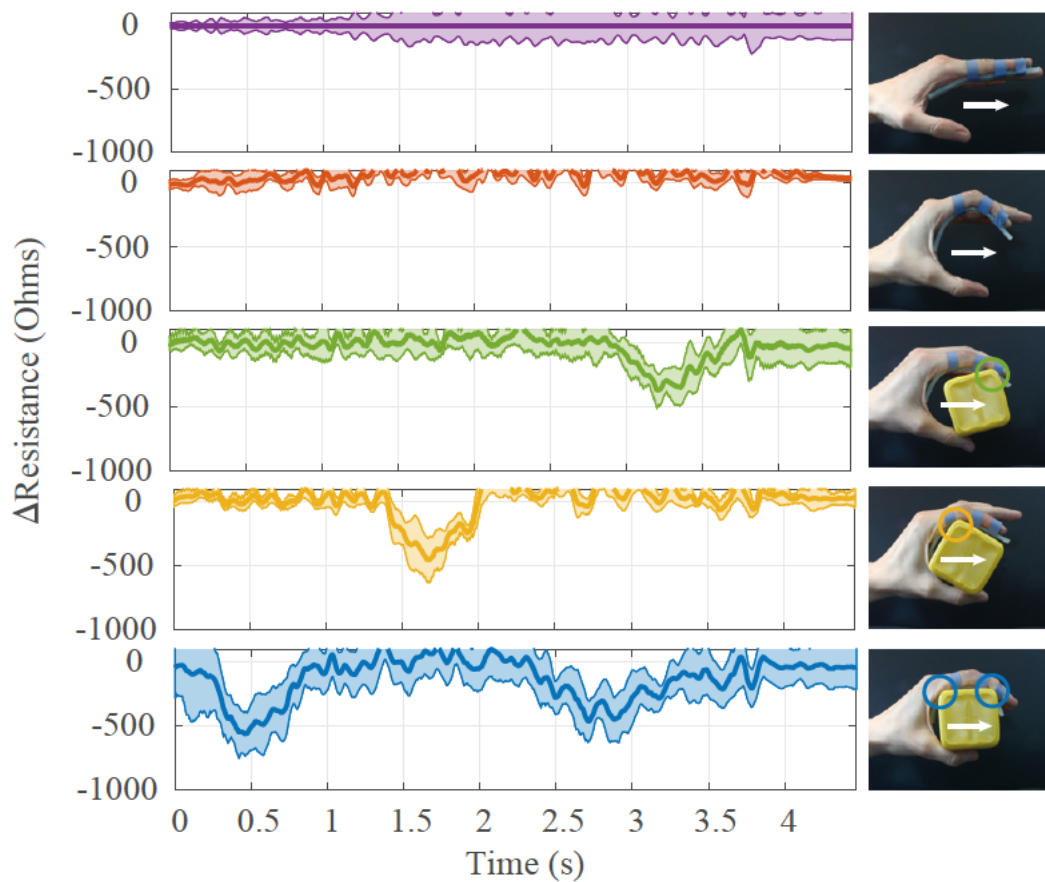


Fig. 5.15 Grip location detection application. A cuboid object was gripped at varying angles. The average and standard deviation across three trials, relative to the baseline of a straight finger, are shown.

knowing the input resistivity signal to retrieve the geometric reconstruction or looking into machine learning models for predictions.

## 5.14 Chapter Summary

Designing information-rich and space-efficient sensors is a key challenge for soft robotics, and crucial for the development of safe soft robots. Sensing and understanding the environmental interactions with a minimal footprint is especially important in the medical context, where portability and unhindered patient/user movement is a priority for decentralised health-care solutions. The ability to tailor and tune the sensing dynamically is also essential for personalised and growing implants. In this chapter, both a soft tunable strain sensor and a pulsating fluidic soft sensor (PFS) are demonstrated. The study on the soft strain sensor is preliminary, but demonstrates the ability to tune the resistance by varying its volume and



ionic concentration and opens up the possibility of creating soft sensors, whose electrical properties could be adjusted dynamically in fluidic soft robots, in order to suit specific tasks. For the more comprehensive study of the PFS, its capacity to determine location, pressure and contact area of press events was shown. Relying on the spatio-temporal resistance changes driven by a pulsating conductive fluid, the sensor demonstrated good repeatability and distinction of single and multiple press events, detecting single indents of sizes greater than 1 cm, forces larger than 2 N, and various locations across the sensor, as well as multiple indents spaced 2 cm apart. Furthermore, the sensor was demonstrated in two applications to detect foot placement and grip location. Overall, the sensors represent an improvement towards minimizing electronic hardware, and cost of the sensing solutions, without sacrificing the richness of the sensing information.



# Chapter 6

## Discussion

### 6.1 Overview

In this thesis, two sensor-actuators have been extensively characterised and demonstrated in a variety of medical applications, alongside the exploration and development of two soft sensors. Resistive-based sensing was used as a low-cost and easy sensing technology and its full sensing potential was explored, trying to maximise the information available in the smallest footprint.

In Chapter 3, a soft sensor-actuator was presented as a building block for a plethora of applications, including as a probe and a robotic implant. The simplicity and versatility of the design, as well as common use of ballooning membranes in soft robots, make it a fundamental study to begin to understand the role and capabilities of ballooning sensor-actuators.

Finally, in Chapter 5, two different sensors were studied, looking at how to maximise information density and tune sensors depending on the application. These investigations emphasised the benefits of dynamic bodied sensors, and moving beyond the current paradigm of sensors with fixed bodies that get placed and left to monitor.

Overall, these chapters have explored how sensing can be integrated within implants, where space is limited, and availability of information is crucial to making informed therapeutic decisions. Compared to the rest of the soft sensor field, this work has furthered the integration of fluidic ionic sensing into two new sensor-actuators, with extensive characterisation allowing further integration beyond the two applications demonstrated. The first two

technical chapters therefore provide some answers to the research questions, going beyond thinking of sensing as merely an addition to the soft robot, but instead integrating it into the body and the actuation itself, thus providing sensing capabilities in a minimum footprint. In comparison to existing research, the last technical chapter explores two novel sensors, and develops the idea of dynamic embodiment of sensing, answering the research questions around new sensing frameworks and maximising sensing information. Using the versatile fluidic medium to drive resistance changes was shown to allow tunability and localisation in a compact size, with minimum external resources. Unlike other groups, that have investigated using different materials or changing the geometry to tune the sensor to a given application, the tunable sensor exploits the easily tunable factors of concentration and volume of ionic solution sensors, to be able to tailor the resistivity of the sensor. The ability to tune two factors gives greater potential to adapt to varying applications and environments, as well as the use of liquid allowing easy reuse and replenishment of a single sensor in applications with different needs. Finally, the pulsating fluidic sensor extends this idea further, with a pulse providing a dynamic excitation and allowing temporal analysis to investigate and diagnose soft robots. This research begins the exploration into the opportunity of having dynamic sensors, going beyond the current static sensing technologies and laying foundations for softer and safer implantable robots.

## **6.2 Limitations and Future Work**

In spite of the promising results demonstrated, a number of limitations still need to be overcome to take this work further. In the following subsections, the key limitations are presented and discussed briefly.

### **6.2.1 Repeatability**

Despite using mainly simple sensor designs with lamination fabrication techniques, the heavily manual process led to poor repeatability of performance. For the soft axially lengthening implant, the glueing step proved to be flawed and caused large variations in the bending angles, and the push placement of the electrodes led to discrepancies in resistance sensing. For the pulsating fluidic sensor, the manual sewing and folding over of the snipped conductive wires caused differences in sensitivity across the length of the sensor. In both cases, exploring more automated fabrication solutions, such as 3D printing, could improve the repeatability and minimise the number of manual steps needed. Furthermore, automated fabrication could also enable more advanced designs, including scaled-down and clinically relevant sized

robots with integrated sensing, to be tested without the fear of compromising repeatability of results.

### **6.2.2 Sensing Drift and Signal-to-Noise Ratio**

Another key limitation of the use of ionic solution as the sensing medium is its sensitivity to changes in temperature and gradual electrolysis over time, leading to drifts in sensing. In addition, the low conductivity of saline solutions reduce the signal-to-noise ratio. While liquid metals have been used in soft sensing, their lack of biocompatibility make them poor choices for implants. Other more biocompatible liquids such as potassium iodide and glycerol, which have shown some promise in being more stable could be explored. One key requirement for the pulsating fluidic sensor though, is the ability of the liquid to achieve laminar flow in the sensing channel, so would need to be chosen carefully to not compromise the working principle for that sensor.

### **6.2.3 Scaleability**

The scale of the sensor still needs further exploration. Although the simpler sensors have shown good integration into small building block actuators or full implants, achieving the more advanced sensing demonstrated in Chapter 5 would need careful consideration of how to integrate such sensing capabilities within the more limited footprint of an implant. Additionally, the extension of the pulsating fluidic sensor to longer and thinner versions to be used on wearables would also require significant changes in the fabrication, with soft printable electronics needing to be explored, in addition to exploring options of moulding soft channels onto flexible electronics. The speed and weight of the scaled up solutions are also worth bearing in mind, with simultaneous segment sensing worth considering to increase speed, and investigating lighter liquids or limiting liquid use to reduce the weight.

### **6.2.4 Integration**

Significant work still needs to be undertaken to integrate the sensors and sensor-actuators developed into resilient soft robots. For improved integration of sensing into the robot body itself, 3D-printed conductive materials should be explored to allow easier integration and more control in the positioning of sensing elements. Additionally, the real-time sensing electronics needed should be considered, to move away from the post-experimental data analysis, and enable immediate sensing information to feed the robot's control algorithm and inform decision-making. More thought would also be required to figure out how to

integrate the sensing, actuation and computational robotic elements into the small footprint the application requires. The greater robot platform, including the fluidic setup and the sensing setup, to drive all of these actuators and sensors also requires more consideration to move away from the current large bench-top setup and move towards a compact and portable solution. Many of the size issues could be addressed by the further development of custom electronic boards, and use of off-the-shelf medical syringe drivers or smaller custom 3D-printed designs, as the current syringe drivers have not been designed optimally. Considering the experimental design as a whole, and not developing the parts separately to be reusable in other experimental setups, would also allow more seamless integration and further size reduction. Finally, the control of the overall system will need to be considered carefully, taking into account the dynamic nature of the body and the peristaltic motion in the gastrointestinal tract in particular, the potential buckling and unintended failures of soft robots under external loading conditions and the optimal stimulation needs for tissue growth. All of these elements can be tested individually with extensive benchmark testing of the robots to improve design and minimise failure, as well as optimising stimulation regime with *in vitro* testing and finally validating the whole system in *in vivo* testing.

[REDACTED]

[REDACTED]

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## 6.3 Conclusion and Outlook

Soft medical robots provide a multitude of challenges from a technological and application specific perspective. The complexity of implants and the tissue engineering context, have driven the exploration of compact sensor-actuator solutions in this thesis, trying to strike a balance between size and richness of information. However, this work has also highlighted some key barriers to the development and wider adoption of soft robots and soft sensors, as well as areas of future exploration.

### 6.3.1 Softer Medical Robots

Soft robots have the potential to revolutionise healthcare, moving away from rigid implants, wearables and more general healthcare products, to devices that seamlessly integrate into the human body. This integration could help pave the way to greater patient care, by enabling more anatomical phenomena to be monitored, as well as for prosthetics and implants, allowing more natural movement and fewer side effects from foreign body response. To achieve this goal more work is needed from medical professionals and engineers working closely together to identify potential applications of soft robots and refine the technology with a stronger awareness and consideration of the clinical requirements. Without this strong collaboration soft robots will struggle to make their way through the research translation

pathway. Beyond the medical focused work, further technology focused work is necessary, to ensure the safety and resilience of the soft robots, before they are let loose in the harsh and demanding environment of the human body.

### **6.3.2 Greater Understanding and Repeatability of Soft Robots**

To this end, more advanced and distributed sensing solutions are needed to push forward the understanding of points of failure in soft robot bodies and begin to prevent the catastrophic bursting failures they are currently susceptible to. This distributed sensing could help reduce overinflation before it becomes a point of failure, as well as generally getting a better picture of the inflation and performance of robots. Gaining a greater understanding of soft robots, could help towards improving repeatability and general usefulness for industry as a whole. Within industry, the improvement of reliability could prove to be a pivotal shift towards widespread adoption across many fields, moving beyond academia and avoiding becoming obsolete. Beyond academia and healthcare, soft robots have extensive potential in applications with more direct interactions with humans, such as the current collaborative robots, where robots could become inherently safe, instead of relying on algorithmic compliance to ensure safety. This characterisation and improved repeatability is likely to come from material scientists and engineers understanding existing and developing new soft materials, as well as manufacturing engineers bringing their knowledge on how to reliably mass produce components. The ability to reliably produce soft technologies will open the door for more creative soft robotic design and easier integration into existing robots, unconstrained by the current time-consuming and unpredictable manufacturing processes.

### **6.3.3 Dynamic-Bodied Soft Sensors**

More generally, from a soft sensor perspective, this thesis has begun to highlight the potential of moving beyond static sensors that are fabricated and placed in a robot or an environment to monitor. Moving instead towards more dynamic bodied sensing, with sensors changing and adapting to situations, the use of pulses could allow engineers to use the diagnostic capabilities of sensing, by having a broader picture of the state of the robot and the ability to query areas of concern. Electronic and material engineers will be essential in providing insights into the choice of materials and mediums to use, as well as suggesting improved methods for signal interpretation. Nevertheless, careful consideration to not unnecessarily complicate systems, and thoughts towards more seamless integration of these sensors is vital. However, once these limitations are overcome there is the possibility of transcending the traditional approach of sensors having fixed sensing mediums that keep sensing properties



over time, opening the door to exploring the rich information density that dynamic bodied sensing could afford. This could not only help sensors to be thought about in a more dynamic way, but also make them more carefully considered as an integral part of the robot body and a vital tool for informed decision-making.

#### **6.3.4 Sustainable and Universal Soft Robotic Components**

Finally, with the greater recent emphasis on recycling and reusability, the consideration of simple sensor-actuator and sensor building blocks, that could be reassembled, reconfigured, or with the sensing medium changed depending on the application, is paramount towards making soft robots more sustainable. The ability to change the body of the sensor could allow one sensing body to be used across multiple applications, tuning the characteristics depending on the application. For the medical applications, this tunability could also help with tailoring solutions to patients, in addition to allowing the sensing to change over the course of the treatment. In the case of a lengthening implant, this could mean the sensor adapting as the implant grows, helping to improve the sensitivity across the range of actuation.

Overall, the future of soft robots and sensors seems bright, as long as the immediate hurdles of reliability can be overcome to allow wider adoption across all industries, while still closely keeping in mind the environmental impact of the developed technologies.



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# Appendix A

## Soft Axially Lengthening Actuators

Extracts, of minor contributions, provided to two other publications, have been included to supplement and provide additional context for the work on the soft axially lengthening implant.

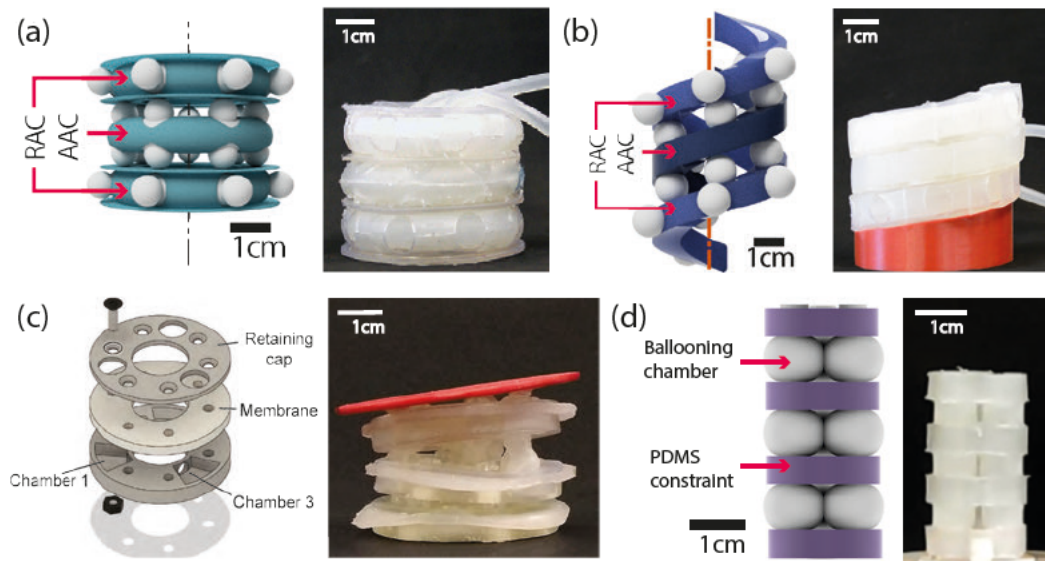


Fig. A.1 Development of soft actuators for tissue regeneration. (a) Toroidal and (b) helical configurations of a fully soft hyperelastic membrane actuator presented in [137] (Used under the Creative Commons License). (c) Hybrid rigid-soft continuum robot, reprinted from [72], ©2021 IEEE. (d) Fully soft anatomically-scaled implant.

## **A.1 Testing the Structural Strength of a Fully Soft Hyperelastic Membrane Actuators in Toroidal and Helical Multifunctional Configurations**

The first work is based on [137] and reprinted under the Creative Commons License. The second work is based on ©2021 IEEE. Reprinted, with permission, from [72].

### **A.1.1 Introduction**

By encoding the capabilities provided by the compliance of the aforementioned requirements into the design of two novel Multimodal Hybrid (M2H) actuators, we introduce the following contributions to this area of research: (1) introduction of the concept of stacked Hyperelastic Ballooning Membrane Actuators (HBMA) realized by 3D arrangements of ballooning membranes; (2) a series of numerical analyses to define the M2H-HBMA design features; (3) proposal of two modular and versatile tubular actuator designs, helical and toroidal, based on HBMA, capable of hyperextensibility and load-bearing; and (4) experimental characterization and validation of the two types of soft actuators.

### **A.1.2 Materials and Methods**

Two different actuators - helical (HA) and toroidal (TA) - were designed and tested. The full details of the fabrication methods undertaken by another lab member are described in [137], and a summary for reference is given below. The Helical Actuator (HA) is shaped out of two Ecoflex 00-30 (Smooth On Inc.) helical chambers made of 8 sections encased into a semi-soft elastic resin exoskeleton to form the Axial (AAC) and Radial (RAC) Actuation Chambers. The Toroidal Actuator (TA) is shaped out of an Ecoflex 00-30 (Smooth On Inc.) toroidal chambers encased in a two-part semi-soft elastic resin exoskeleton to form the AAC and RAC. For both designs, stacks of AAC and RAC are interlayered to provide both axial and radial growth. Following the concept of stacked balloons, the HA and TA axially extend by expanding balloons out of their AAC, displacing the stacked RACs. Similarly, when pressurized, the membranes in the HA's and TA's RACs expand and shape balloons in the radial direction.

### **A.1.3 Results**

TA's AAC shows the best performance of the two M2H-HBMAs, with over 150% maximum extension for both the preloaded and post-loaded conditions compared to a maximum

extension of 115% for the HA's AAC (Fig. A.2(c)). All percentages of extension presented throughout this section were calculated using the formula:  $\frac{L-L_0}{L_0} * 100$ , where  $L$  is final length and  $L_0$  is initial length of the AAC.

TA's AAC is also capable of greater extension, when preloaded with extensions of more than 150% as compared to less than 30% for the HA's AAC. The HA's and TA's AACs both perform better when post-loaded, showing and maintaining greater extension under heavy loads, demonstrating extensions of 60% and 160% with 1 kg loads respectively compared to less than 20% extension for both when preloaded with 1 kg. Finally, the extension of the TA's AAC is considerably more consistent for the post-loaded condition showing an average drop of around 15% extension from 0 kg load to 1 kg load compared to around 160% for the preloaded condition.

Fig. A.2 (d,e) shows the extension of the HA's and TA's AAC under gradual pressurization to 24 kPa. Both the HA's and TA's AAC require pressures of around 15 kPa to begin extending under no load conditions. For the TA's AAC, as the load increases, the input pressure needed to begin extension also increases, from around 15 kPa under no load to around 20 kPa for 0.4 and 0.6 kg loads. A similar behaviour is observed for the HA's AAC, although much less distinct due to an overall poorer extension.

Under the preloaded condition, the PET sheet, which represents a structural scaffold of the organ, has little impact on the axial extension, as the loading already restricts the ballooning in all directions. In the post-loaded condition, due to the AAC-only inflation, there is limited contact between the balloons and the PET sheet in the pressurizing stage. In the loading phase, the PET sheet prevents the compression of the AAC balloons in the radial direction due to the load, and consequently the potential burst of the AAC. While the PET sheet does not model the mechanical properties of a tubular tissue, the sheet plays a generic role in geometrically constraining the actuator.

#### A.1.4 Discussion

The M2H-HBMAs showed the ability to extend when loaded before and after pressurization with up to 1 kg (Fig. A.2 (c)). Both the HA's and the TA's AAC were able to withstand greater loads whilst maintaining extension levels when loaded after pressurization, showing good structural strength once pressurized. When pressurizing under load, the TA's AAC performed better, showing greater active motion under external loads. The difference in extension when preloaded could be from a more uniform distribution of the load across the balloons of the actuator in the TA's AAC or from the different cross-sectional shape of membrane, with the load only being in contact with the top of the balloon for the circular cross-section in the TA's AAC compared to the whole area for the square cross-section in the HA's AAC. The extension

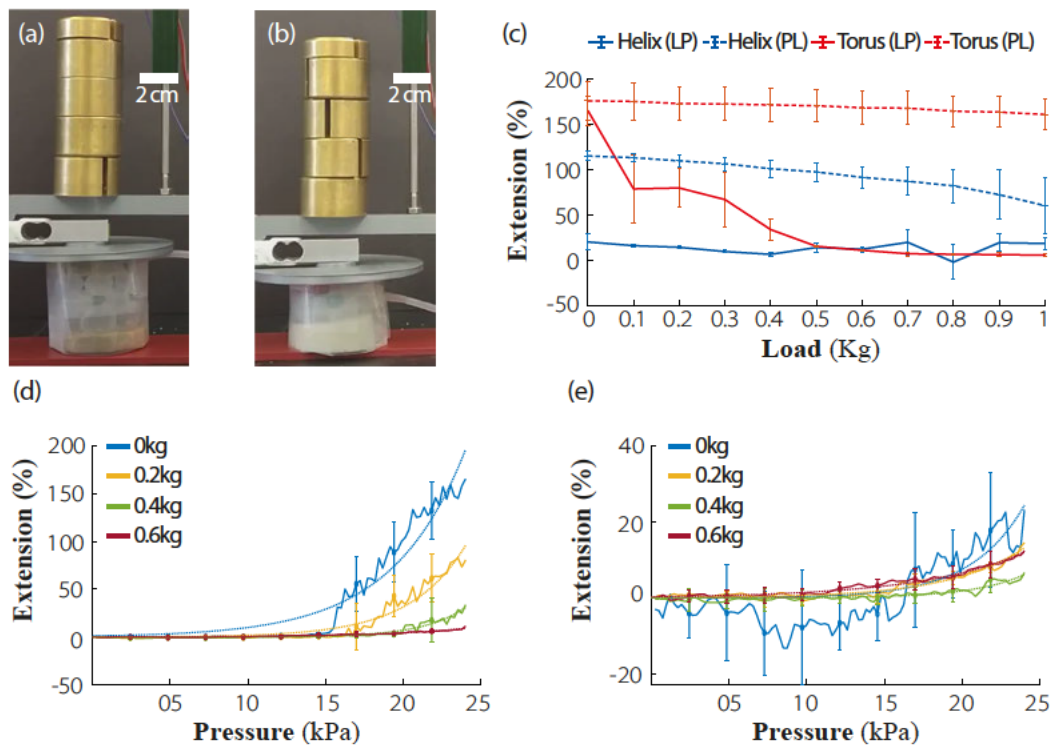


Fig. A.2 Schematic of the inflated SA.

of the actuator could also be increased through greater pressurization when constrained and under heavier loads, as the level of required input pressure to induce extension was shown to be greater for heavier loads (Fig. A.2 (d, e)). Given that the presented setup is a geometric-only representation of a tubular organ, further experiments are needed. The effects of a radial constraint for the load bearing and force exerting capabilities of the TA's and HA's AACs need to be evaluated using tubular constraints that have the mechanical compliance, friction coefficients and clinical specification according to the medical condition being targeted for treatment. Finally, in the clinical setting the traction forces exerted on the tissue will also depend on the attachment methods used as an interface between the actuator and the organ [136].

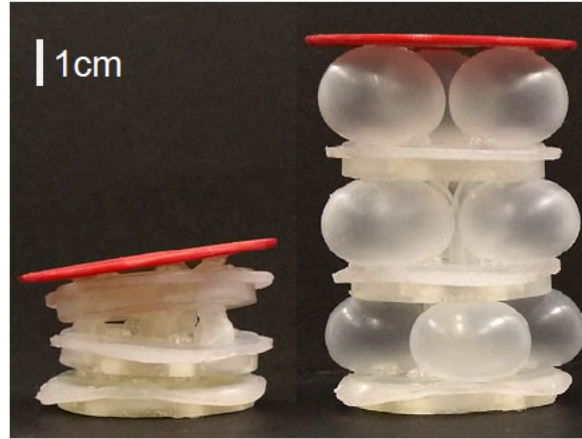


Fig. A.3 Schematic of the fully deflated and fully inflated stacked actuator.

## A.2 Determining Range of Motion of a Hybrid Soft-Rigid Continuum Robot

### A.2.1 Introduction

Extending on our group's previous soft implantable actuators capable of elongating 36.3% [138], we propose the design and model of a highly extensible and tough actuator: the Hyperelastic Ballooning Membrane Actuator (HBMA). The contributions of this paper include: (1) a model of pressure, membrane height and object interaction of a hyperelastic ballooning membrane controllable by the applied voltage to a micro-compressor (2) easy fabrication of a hyperelastic ballooning membrane actuator by leveraging a hybrid material approach (3) the proposal of a flat actuator which is both highly extensible and compressible.

### A.2.2 Materials and Methods

A hybrid soft-rigid continuum robot was designed, developed and modelled by another lab member, with the full fabrication detailed in [72] and a summary provided below. The actuator consists of a thin hyperelastic ballooning membrane sandwiched between two SLA 3D-printed parts (Form 2). The membrane is cast of Ecoflex 00-30 (Smooth On Inc.) cured at room temperature and further post-cured in the oven. The bottom SLA-printed part accommodates three chambers, on which the membrane is placed, and then secured with the top SLA-printed part, which screws to the base. Each of the three ballooning membranes can be actuated individually to allow the overall actuator three degrees of freedom. In the final application, two layers of three ballooning membranes are stacked, and connected with cyanoacrylate to create the hybrid soft-rigid continuum soft robot.



Table A.1 Average angle and displacement between two layers

| Number of HBMA inflated | angle (deg) | length (mm) | extension (%) |
|-------------------------|-------------|-------------|---------------|
| 0                       | 2.3         | 9.6         | 0             |
| 1                       | 28.9        | 18.0        | 88            |
| 2                       | 28.5        | 23.0        | 140           |
| 3                       | 1.3         | 26.7        | 179           |

### A.2.3 Results

Fig. A.4 shows how the hyperelastic ballooning membrane actuators (HBMA) can be used in a hybrid soft-rigid continuum soft robot. The robot is actuated sequentially generating the displacement and rotation of the layers. Without reaching the limit (which would mean breaking the membrane) the average angle between one layer and the next one, as well as the displacement are summarized in Table A.1. The maximum displacement and rotation of the robot's end-effector can be further increased by stacking more layers.

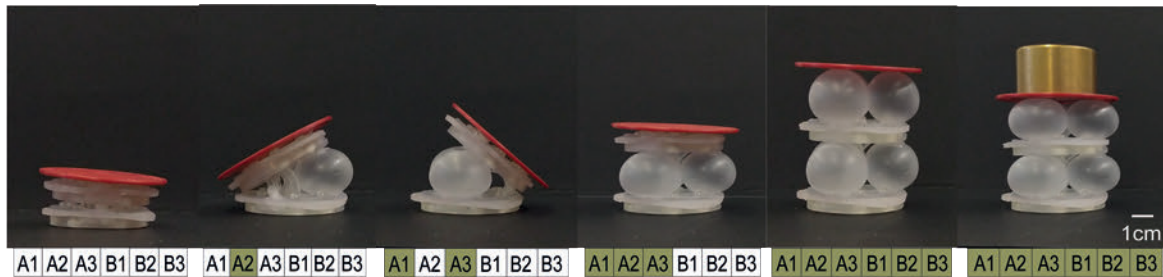


Fig. A.4 Photos of the stacked actuator showing full range of motion and structural strength capabilities. Balloons of the stacked actuator are simultaneously pressurised and individually pressurised to show the variety of angles and growth achievable for the stacked actuator. The stacked actuator is also showed fully inflated with a loading of 200g.

### A.2.4 Discussion

Fig. A.4 also shows that the model previously presented can be used to estimate the height of each HBMA sequentially actuated. From the estimated displacement, the position and orientation of the last layer could be computed using a forward kinematic approach for continuum robots [80].



# **Appendix B**

## **Evaluation of Tissue Attachment Interfaces for Soft Implantable Devices**

In this appendix, work investigating two different attachment interfaces for a soft robotic implant are presented. The first interfacing method consists of attachment using a fibrinogen matrix, naturally adhering to the tissue, which is glued to a soft attachment. In contrast, the second method consists of a fabric interface embedded into the soft attachment, which is subsequently sutured to the tissue.

The work was unsuccessfully submitted as a short paper to IEEE Transactions on Medical Robotics and Bionics and is presented below in single column format. The paper was collaboratively written with Colleen Hood as co-first authors, and Eduardo Perez-Guagnelli, Abigail F. Smith, Nicolas Herzig, and Dana D. Damian.

### **B.1 Abstract**

Soft materials have been shown to reduce fibrotic response of soft tissues, as well as to comply better to restricted areas inside the body and have therefore shown great promise for soft implantable robots. However, the attachment methods between soft tissues and soft implants have yet to be fully investigated. Challenges in attachment methods between soft robots and biological tissue include finding optimal material combinations to reduce mechanical stresses at the tissue-robot interface and guaranteeing the necessary adhesive strength to both soft materials and tissues, in order to transfer the required mechanical work through this interface with little damage. In this paper, we compare the use of fibrinogen adhesives and sutures to interface between soft tissues and attachment connectors. We present the force and stiffness results for tests carried out on rabbit and porcine esophagi. The average

rupture force using fibrinogen attachments was 2.4N for the rabbit esophagi and 2.8N for the porcine esophagus, compared to forces greater than 4N being withstood across all of the suture trials. This study highlights the importance of the attachment interface for soft robotic implants and gives an insight into the trade off between maximum force that can be applied and tissue/device safety.

## B.2 Introduction

Soft robots have received increasing interest in healthcare and clinical settings, including minimally invasive surgery [158], wearables and implants [139]. Their mechanical compliance makes them ideal for implantation and conforming to restricted spaces inside the body, while causing minimal damage and foreign body response [119] [41].

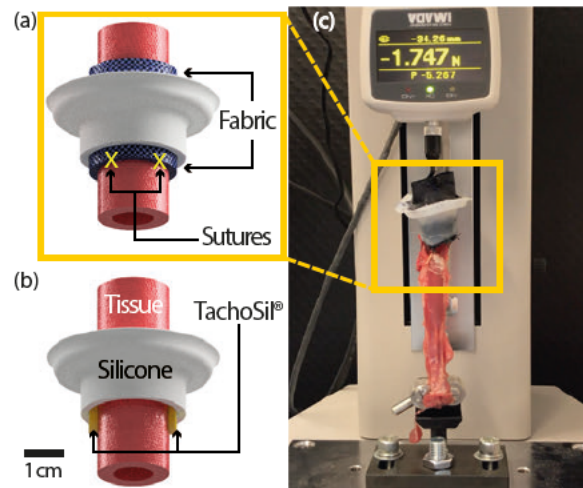


Fig. B.1 The tissue attachments for soft implantable devices. (a) Attachment tissue-implant using sutures and (b) TachoSil<sup>®</sup> membrane. (c) The experimental setup.

There are various methods available to secure implants to target tissues. In most cases, sutures are used to attach implants to delicate soft tissues. Other devices such as over-the-scope clips [91] or the BRAVO capsule [19] also rely on puncturing or pinching tissue to attach themselves. This invasive puncturing can lead to damage of the tissue, posing higher infection risks than adhesive alternatives and resulting in potentially longer recovery periods [113]. Alternative attachments are through medical adhesives [101]. Fibrinogen sealants and adhesives use the concluding stage of the coagulation cascade to form an adhesive cross-linked fibrin clot [186]. These fibrinogen sealants are traditionally used to seal surgical wounds and reduce internal bleeding.

One challenge that remains to be solved is reducing the impact of attaching implants to soft tissues. Due to the difference between the material properties of implants and native tissue, fibrotic encapsulation may occur around implant. Designing attachment methods that reduce the stresses at material interfaces, while also maintaining adhesion strength is essential to reduce fibrosis.

Another challenge is the ability to transfer the required mechanical work through this interface. This issue is of particular importance for implants designed to grow tissue through mechanostimulation as this process requires transmission of forces between the two.

In a recent work, our group introduced a soft implant for tubular tissue regeneration [139]. We investigated the effect of stiff and semi-stiff attachments that connect the soft robotic implant to the host organ when forces are applied to the tissue [135]. We observed that the semi-stiff attachments resulted in a smoother transfer of applied force; an important feature, as force peaks could lead to tearing of the tissue.

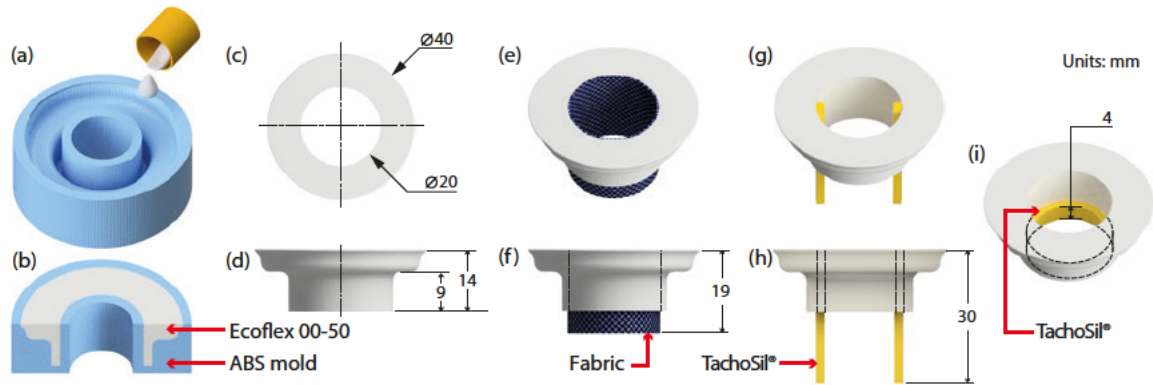


Fig. B.2 Fabrication process for the soft attachments. (a) Mold for casting of the soft attachments. (b) Cross-sectional view of the soft attachment inside the mold. (c) General top and (d) side dimensions of the soft attachments. (e) The attachment method using sutures. (f) General dimensions of the attachment with the embedded fabric for suturing. (g) Method using an FM: (h) General dimensions of the attachment with the glued TachoSil® strips for rabbit and (i) pig tissue.

In this study, we extend our previous work investigating a suitable interface between soft implants and biological tissues by presenting a comparative study of tissue attachment interfaces for soft robotics implants. These interfaces are based on two methods: (1) suturing method with surgical needle and wire (Fig. B.1(a)), (2) attachment using fibrinogen matrix (FM) (Fig. B.1(b)). The two candidate solutions are evaluated based on their rupture resistance measured by a tensile testing machine (Fig. B.1(c)).

## **B.3 Materials and methods**

### **B.3.1 Soft Tissue Attachments Fabrication**

Molds were fabricated via rapid prototyping out of ABS material (Prusa 3D printer) to cast the soft attachments (Fig. B.2(a)). We mixed and degassed (Thinky Mixer ARE-250) silicone rubber (Ecoflex 00-50, Smooth-On Inc.) that we then poured into the ABS molds (Fig. B.2(b)). The silicone parts were left to cure at room temperature for three hours, followed by a post-curing process of two hours at 80° and then one hour at 100° using an oven (SciQuip®). For the attachment method using an FM, two strips of 3 by 0.4cm of TachoSil® were glued to the silicone rubber using cyanoacrylate adhesive (Fig. B.2(g-h)). For the method using sutures, we embedded polyester fabric into the silicone attachment using uncured Ecoflex 00-50 (Fig. B.2(e-f)). It was then left to cure at room temperature for three hours.

### **B.3.2 Attachment to tissue procedure**

To attach the FM to the tissue, we pressed the two together for 3 minutes to bond them, around the circumference of the larger porcine esophagus and longitudinally along the rabbit esophagi (Fig. B.2(g-i)). One sample of porcine tissue and three samples of New Zealand white rabbit tissue were used in this study. These numbers were limited by the tissue that was available. Each tissue sample was first tested with the FM and then with the sutures.

To attach the polyester fabric to the tissue, we used four simple interrupted sutures made with black braided non-absorbable suture silk and a X-1 CONV 22 mm needle (MERSILK® kit). This suturing technique is commonly used in surgery to close wounds. Fig. B.1(a) and Fig. B.1(b) show a schematic of the TAs with FM and suture attachments respectively. The extra layer of fabric was added for the suturing method. We found that when directly suturing the silicone rubber, the suturing wires were tearing the silicone, even for small applied shear forces.

### **B.3.3 Tensile Experiment procedure**

We used the EMX-5N IMADA tensile test machine to measure the force exerted on the tissue and TA along with the displacement.

The porcine and rabbit esophagi bonded to the TAs were placed on the tensile test machine. The soft cap was used in replacement of the soft robot and was hooked to the top of the tensile testing machine, the ex vivo tissues were clamped to the lower part of the machine.

The tissue was then stretched at a speed of 50mm/min until either the failure of the attachment or the force sensor limit (5N) was reached. During the trial, the displacement of the tensile test machine and the traction force were acquired at 1kHz. Ex vivo experimentation was considered the only suitable type of experimentation where rupture force can be investigated with animal tissue. This is due to the ethical implications of rupturing tissue in vivo, which would cause significant harm to the animal.

## B.4 Results

Fig. B.3(a) shows the increasing force as the tissue was stretched before failure for both interfaces for one trial of each tissue. It should be noted that, in this figure, the force signals have been filtered (with a moving average filter) and down-sampled to reduce the measurement noise.

In Fig. B.3(b), the rupture forces for the three trials on the rabbit esophagi and the single trial on the porcine esophagus, conducted for the two different attachment methods, are shown. The rupture force was defined as the maximum force achieved before the attachment failed. The FM withstood a maximum of 3N and 4N for the rabbit and porcine trials respectively. In each trial the FM interface failed, leaving the mesh adhered to the attachment. For the simple interrupted sutures, a force greater than 4N was withstood in all of the trials, with the suturing method reaching the 5N limit of the load cell for one of the rabbit esophagi trials and the porcine trial. For the trial where the 5N limit was not reached, the tissue ripped before the sutures failed. It can be noticed that in Fig. B.3(a), the illustrated trial for the porcine tissue is where the suture failed.

Additionally to the maximum forces achieved, some sudden drops of force can be observed during the tests. To highlight these drops, the force gradient  $\frac{\partial F}{\partial x}$  has been computed and shown in Fig. B.4. This gradient can be seen as the equivalent stiffness of the TA in series with the tissue sample. It can be seen that this stiffness tends to increase slowly with the elongation, but reaches some local plateaus before the gradient finally drops at the rupture. The local force drops are also highlighted in Fig. B.4 by the negative peaks in the gradient.

## B.5 Discussion

In a previous study [135], we have shown that a 2.5N force was enough to grow tissue by mechanotherapy with a robotic implant. The current study demonstrates not only that the classic suturing approach easily withstands this force, but also that with the TachoSil®, the rupture force of the attachment was on average 2.4N for the rabbit esophagi and 2.8N for

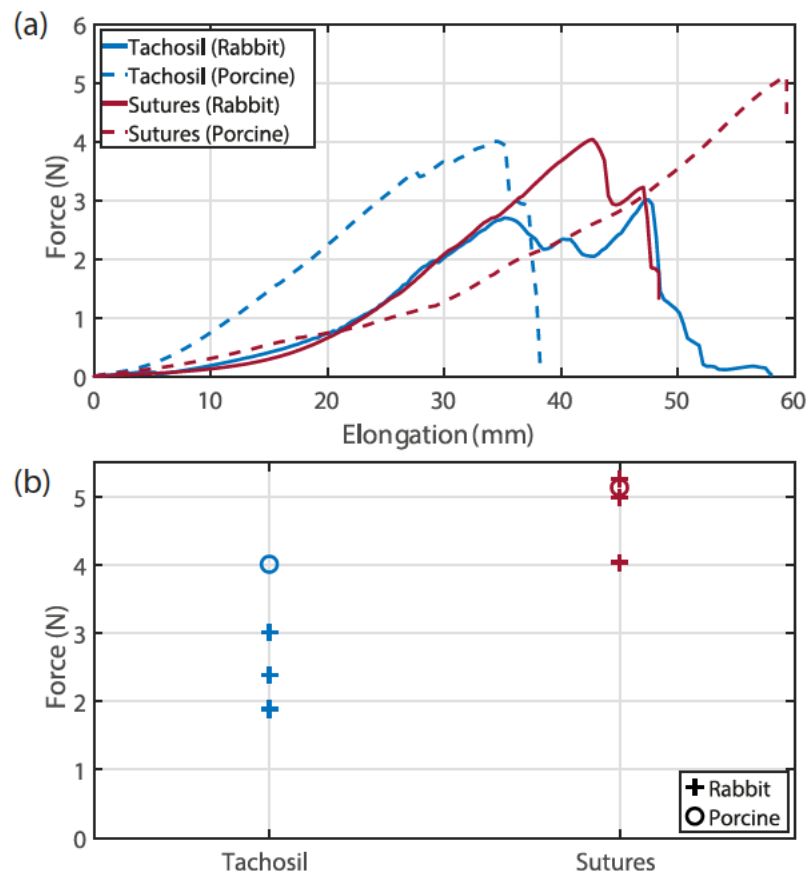


Fig. B.3 Forces measured during elongation and at rupture. (a) Force response to the elongation of the tissue for both attachment methods and for a single trial of both the rabbit and porcine esophagi. (b) Rupture force, maximum force applied before attachment failed, of the attachments for both rabbit and porcine esophagi.

the porcine esophagus. These results suggest that an FM could be used to replace traditional suturing methods to attach soft robots to tissue. The difference in the rupture force between the rabbit and the porcine tissue can be explained by the difference in the size of the tissues or the placement of the FM. Indeed, the porcine tissue has a larger diameter which allows for a more uniform contact between the FM and the tissue.

The tissue morphology also influenced the direction in which the FM was attached and differed for the two types of tissue samples, as shown in Fig. B.2(h-i). Although the area of FM used to attach the TA to the tissue was consistent throughout all the experiments, the attachment technique varied. This could have affected the rupture force, as although the forces are being exerted in the same direction, the attachments for the rabbit (Fig. B.2(g-h)) are likely to induce more localized stresses. Furthermore, the FM was secured to the tissue over a larger area than the sutures. The sutures were attached at multiple localized points. It



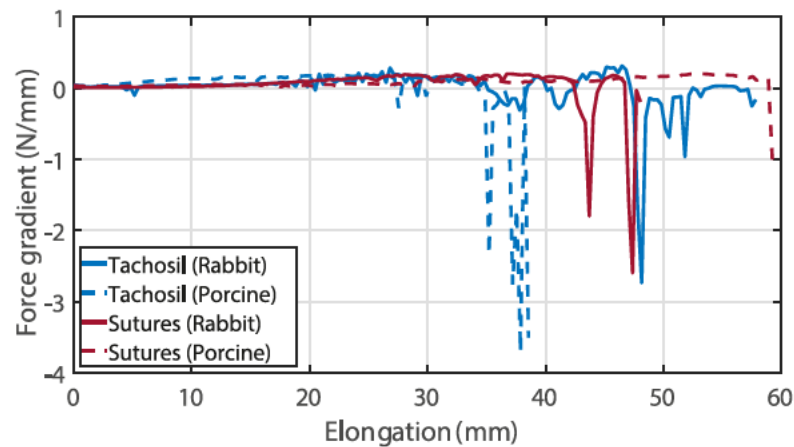


Fig. B.4 Gradient of force during elongation of the tissue. The negative peaks represent local force drops indicative of rupture.

can be hypothesized that the sutures generate greater localized forces in the tissue. These forces are also dependent on the tightness of the suture which is difficult to control as it is dependant on the surgical practice [188]. These variable localized forces can alter the surrounding cell morphology or damage the tissue [174].

The observation of the samples after failure shows that the rupture with the FM attachment occurred when the latter was peeled off the esophageal tissue. Moreover, this happened without apparent damage at the tissue level. On the other hand, for the suture attachment, excessive forces applied may result in the attachment not failing, but tearing damage on the tissue nearby the suture. The peeling behavior of the FM may be seen as a mechanical fuse that peels off in case excessive forces are applied. This behavior seems interesting for real case scenarios in order to prevent involuntary tissue damage and to protect the soft implantable device and raises the possibility of tuning the strength of a TA interface depending on its function.

Stiffness has been shown as a mechanical cue for cell signalling [195] and therefore its variation is likely to impact cell proliferation [131] and Extracellular Matrix production [182]. In Fig. B.4, we can see a slight upwards trend of all samples in the force gradient, regarded as the equivalent stiffness. This trend is due to the non linear elasticity of tissue [57]. The purpose of the attachment is not to induce cellular response, but to secure the implant in vivo. A stiffness that is more homogeneous and compliant to the tissue is therefore less likely to cause adverse effects, such as fibrosis. In this study, there is no clear difference in overall stiffness between the FM and the sutures.

The sudden drops seen in Fig. B.4 could be explained by two phenomena. Firstly, this may be the result of a partial rupture of the attachments. The second is a possible tearing or

plastic deformation of the tissue. However, in real case applications, these scenarios should be detected and avoided. Future work will involve investigating and further distinguishing these two types of force drops, in order to develop dedicated fault tolerant strategies for soft robotic implants [9].

There were some limitations in the materials and methods used. With regards to the tissue, each sample was first tested with an FM and then with sutures, given the limited availability. The FM detached from the tissue before any damage to the tissue was visible, but any plastic deformation induced by the initial tensile testing with the FM, may have impacted the results of the latter tests with sutures. Some materials used in this study (cyanoacrylate adhesive, polyester fabric and Ecoflex 00-50) are not graded as biocompatible. Non-biocompatible materials are generally cheaper than their biocompatible alternatives and share similar physical properties. They can therefore be substituted in for future experimentation. Finally, the study outlined in this paper was entirely *ex vivo* and was not conducted under sterile conditions. For future *ex vivo* and *in vivo* experimentation, the tissue attachments can be fabricated under sterile conditions in a biosafety cabinet, to improve the condition of the biological samples.

## B.6 Conclusion

For the first time, a comparative study is presented for tissue attachment interfaces for soft robotic implants. On the one hand, a weaker TA interface, such as that found with TachoSil<sup>®</sup>, can be used as a safety fuse to ensure safe mechanical stimulation that detaches from the tissue if forces exceed around 2.4N. On the other hand, a stronger TA interface, such as that found with the sutures, has been shown to withstand a higher force, which may be useful when more pronounced actions are needed to stimulate the tissue, at the risk of tearing tissue and causing damage.

One clear next step of this study is to conduct further trials on fresh biological tissue samples. Due to limited tissue availability, the method was only tried on three rabbit esophagi and one porcine, so future work would include more trials on tissue. Additionally, given the long term goal of implants in humans, conducting experiments on tissue closer to the human esophagus morphology is vital. Performing trials *in vivo* or *in vitro* is also crucial to understand the behavior of the attachment method during tissue growth, as *ex vivo* tissue may have slightly different mechanical properties due to cadaveric rigidity.

Additionally, the attachment methods need refinement and further study. While the FM shows promise as an attachment method, future work still needs to be conducted to understand the impact of the contact surface and long term stress on the rupture force and



overall condition of the interface. Methods to embed the FM directly in the TA to prevent use of non-biocompatible glues also need to be investigated. With progression in surgical techniques, we could also envisage changing the fabric used for the suture attachment to a tissue graft from the same patient.

Finally, conducting the experiment again with biocompatible materials would give insight as to whether non-biocompatible materials are viable substitutes for preliminary testing. In line with changing to biocompatible materials, sterilisation techniques can also be tuned and implemented during the fabrication process.



## Appendix C

# Pneumatic Kitchen: Hand-Holding Gloves

In this appendix, work in collaboration with academics Kaisu Koski and Roland van Dieren-donck from the Lab4Living at Sheffield Hallam University is presented. The work was exhibited at the Caring Futures exhibition at Sølverget Galleri, Stavanger, Norway from September 17 to December 18, 2022, as well as at the Some Bodies: Subject, Object and Identity in Nursing and Art at Opalka Gallery in Albany, NY, from November. 11, 2022 through to January 7, 2023.

My contribution, in collaboration with an MSc student and assistance from other members in SBRL, consisted of assisting with the fabrication of the pneumatic pockets for the glove, to mimic the hand-holding motion, as well as the soldering and testing of the pneumatic circuitry (Fig. C.1)

A summary of the ambitions of the project is presented below: ‘These inflatable glove prototypes belong to a speculative empathic clothing line, simulating two different styles of handholding by fillable airbags and a pneumatic circuit. According to a pre-programmed rhythm and intensity, one of the gloves simulates a hand squeeze, and the other forms a hand sandwich. Hand sandwich is a gesture in which someone takes your hand between both of their hands. The term originates from a medical school classroom, where students learn to express empathy toward the patient. The project is a collaboration between artists and engineers, and the cross-fertilization of their disciplinary viewpoints took place through the artists’ residence in an engineering lab.’

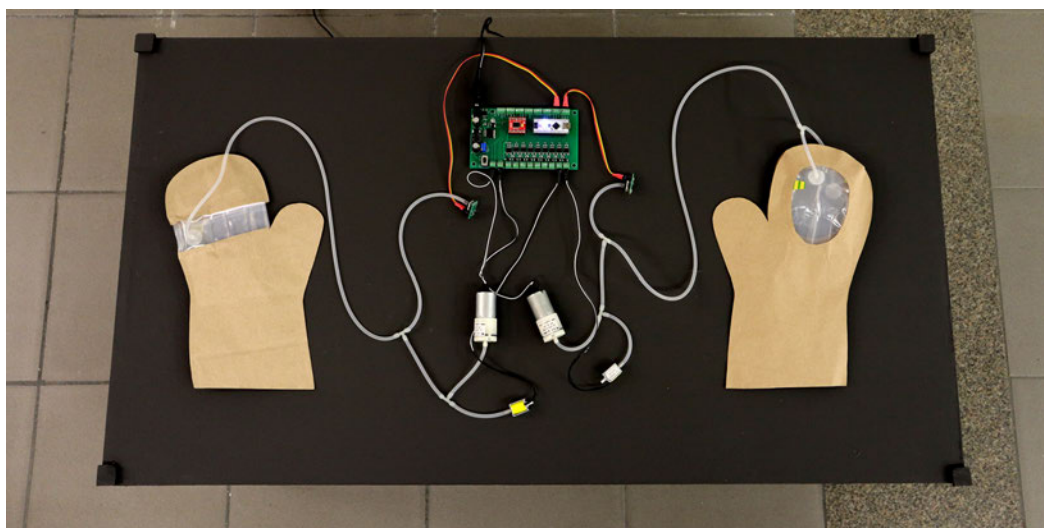


Fig. C.1 Photo of hand-holding gloves with pneumatic circuit

# **Appendix D**

## **Paper on SoPHIA Implant**

This publication features contributions primarily developed prior to the PhD and is included for reference. The work conducted at the start of the PhD mainly included the co-development of the analytical model and its experimental validation.

The work is based on ©2020 IEEE. Reprinted, with permission, from [138].

# Characterization, Simulation and Control of a Soft Helical Pneumatic Implantable Robot for Tissue Regeneration

Eduardo Perez-Guagnelli<sup>1</sup>, Joanna Jones, Ahmet H. Tokel, Nicolas Herzig<sup>1</sup>, Bryn Jones<sup>2</sup>, Shuhei Miyashita<sup>3</sup>, and Dana D. Damian

**Abstract**—Therapies for tissue repair and regeneration have remained sub-optimal, with limited approaches investigated to improve their effectiveness, dynamic and control response. We introduce a Soft Pneumatic Helically-Interlayered Actuator (SoPHIA) for tissue repair and regeneration of tubular tissues. The actuator features shape configurability in two and three dimensions for minimal or non-invasive *in vivo* implantation; multi-modal therapy to apply mechanical stimulation axially and radially, in accordance to the anatomy of tubular tissues; and anti-buckling structural strength. We present a model and characteristics of this soft actuator. SoPHIA reaches up to 36.3% of elongation with respect to its initial height and up to 7 N of force when pressurized at 38 kPa against anatomically-realistic spatial constraints. Furthermore, we introduce the capabilities of a physical *in vivo* simulator of biological tissue stiffness and growth, for the evaluation of the soft actuator in physiologically-relevant conditions. Lastly, we propose a model-based multi-stage control of the axial elongation of the actuator according to the tissue's physiological response. SoPHIA has the potential to reduce the invasiveness of surgical interventions and increase the effectiveness in growing tissue due to its mechanically compliant, configurable and multi-modal design.

**Index Terms**—Implants, medical robotics, soft robotics.

## I. INTRODUCTION

THE IMPACT of soft robots has been seen throughout the medical field, e.g., in assistive technologies and rehabilitation [1], minimally invasive surgery [2], implants [3], and wearables [4]. One of the main advantages of soft robots is their compliance, enabling safe interaction with the human body and thus, increasing the wearability of technologies for the treatment of various clinical conditions.

Tissue repair, in particular, can benefit from the characteristics of soft robots, such as gentle dexterous handling, palpation, anatomical and functional support. Advanced surgical tools, e.g., Da Vinci robots, have demonstrated the benefit

of added accuracy and minimizing invasiveness to the surgical procedures for tissue repair. However, they require surgeon supervision during the surgery, but provide no further control of the tissue repair beyond the surgical intervention. Alternatively, tissue engineering aims to restore the structure and function of a tissue by stimulating cells to proliferate on scaffolds using chemical growth factors [5]. Despite advances in structural tissue regeneration, tissue engineering faces challenges such as lack of vascularity in new tissue, poor mechanical compatibility, and lack of control of the regeneration process after the scaffold implantation [6].

Mechanotherapy has been found to have assistive and therapeutic effects for a variety of medical conditions [9], including tactile sensory restoration [10], wound healing [11], regeneration of skeletal muscle and esophageal tissue [12], bone growth and skin grafts [13], [14].

Robotic implants are a new medical technology with the potential to unify the advantages of mechanical stimulation to tissue repair and regeneration, by exploiting the cells' intrinsic proliferation mechanisms. They also have the potential to engineer and control the process of tissue repair, brought by medical devices, robots and surgical assistance, during the entire duration of treatment via remote communication [15]. These regenerative implants may be deployed inside the body and be mounted on the target tissue. There, they will exert controlled and gentle forces, and elongation on the tissue to induce regeneration and healing.

An example of potential therapies in which robotic implants may be of use is the regeneration of the gastrointestinal (GI) tract. This type of regeneration is required for conditions such as long-gap esophageal atresia (LGEA) (Fig. 1(a)) or short bowel syndrome (SBS), congenital conditions in which more than two-thirds of these organs may be missing [16], and for which current treatments prove suboptimal with high mortality rates [17]. Our group recently introduced a robotic implant that was able to induce growth of esophageal tissue using mechanostimulation and showed 77% of tissue growth over nine days [12]. This rigid implant was mounted on the esophagus and gently pulled the tissue axially. It was able to apply traction forces in the range of 1-2.5 N and displaced up to 5 cm of tissue [3].

Despite the potential of these robotic implants, the *in vivo* studies also revealed a range of unanticipated challenges due to the fixed design of the implant operating long-term in a harsh

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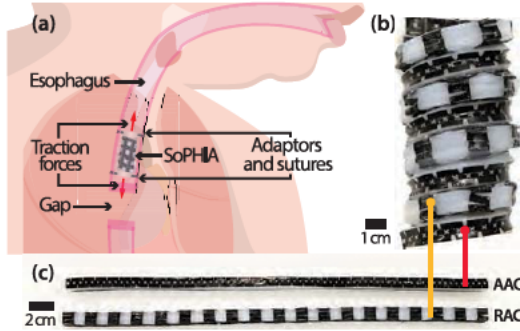


Fig. 1. The soft helical actuator for mechanostimulation-based tissue regeneration. (a) The envisaged implantation of SoPHIA inside the esophagus to treat the long-gap condition. It may be attached to the tissue by using sutures or adaptors [7]. (b) A view of the helical actuator made by configuring two actuation chambers. (c) The two chambers, axial (AAC) and radial (RAC) before the helical configuration to stimulate the tissue axially and radially respectively [8].

*in vivo* environment. Given the interaction between the rigid implant and the tissue, we ascertained that tissue scar level was significant, 37% of the new tissue, because of collagen formation.

The derived requirements for mechanostimulation-based robotic implants for tissue regeneration are: (1) mechanical compliance to reduce inflammation [18]; (2) minimally invasive implantation by surgeons in various parts of the GI tract; (3) a design that allows hyperelastic linear deformation while securing anti-buckling structural strength to resist considerable loading; (4) safe control of the delivery of mechanical stimulation.

To address these requirements, we introduced the concept and fabrication of a Soft Pneumatic Helically-Interlayered Actuator (SoPHIA). This multi-modal soft actuator is composed of two pneumatic chambers coiling together into a tubular implant for mechanotherapy tissue regeneration (Fig. 1(b-c)) [8]. Due to its helical arrangement and reinforced walls, the soft helical actuator is capable of both axial elongation and radial expansion to stimulate the growth and function of the longitudinal and circular muscles of the GI tract. In this work, we present a holistic characterization of the soft actuator's capabilities in the axial, radial, and combined actuation mode, a simulation using a novel *in vivo* tissue growth simulator, as well as the control of SoPHIA.

## II. RELATED WORK AND DESIGN REQUIREMENTS

The clinical requirements mentioned in the previous sections translate into a number of technical requirements. First, the actuator should be soft to mechanically comply with the target tissue. Recent studies in material engineering and tissue regeneration suggest that mechanically compliant implants [18] and the application of mechanotherapy may reduce the inflammatory response of the body [19]. Soft-matter robotic implants would thus be a suitable choice for mechanotherapy-based tissue repair.

Second, the actuator should be modular, to facilitate implantation, and configurable, to allow its mounting on tubular tissues in a non-invasive or minimally invasive manner. Soft actuator research has generally focused on pre-programmed axial lengthening and bending actuation based on fluidic control, for navigation in surgical applications or for exoskeletons [20]. Advances have been made towards modular soft robots for assistive applications [21]. Such devices are highly desirable for non-invasive surgery and implantation as well, yet have not been tested for their reliable extension under sustained loading conditions, as required for mechanostimulation. More recent work has also explored the strength of these actuators with programmable composites, demonstrating the exertion of axial forces to lift up to 1kg after being pressurized at 23.8 kPa, and the expansion of more than 200% their original size [22]. However, the actuator has a large volume due to its origami creases, which is not desirable for an implant that seemingly coats a tubular tissue. While soft medical devices and implants have been recently proposed [23], more work needs to be done to reduce the implantation invasiveness.

Third, the actuator should elongate with the growing tissue without buckling due to load from the tissue or the body's spatial constraints. As soft actuators' directional expansion needs to be increased, to support large tissue reconstruction, it is critical to have actuators that maintain structural strength and do not buckle under load. The compliance of soft robots is limiting in this regard, making it challenging for them to withstand forces from their environments [24].

Helical or coiled structures are widely found, in both biology and engineering, due to their ability to handle tension and compression [25], or to provide enhanced maneuverability and stability, for example to endoscopic instruments [26].

When used in soft robotic systems, they also provide increased dexterity, more efficient workspaces by using routed tendons [27], enhanced area of contact and stability when acting as a gripper [28]. Nevertheless, scarce attention has been put into their investigation.

Fourth, a control of the actuator similar to the current clinical treatment is needed. In general, it is difficult to develop accurate models and control of soft actuators [29] due to the diversity in hyperelastic-based material constituents. Recent research has developed a variety of new methods for control, capable of dealing with the increased dimensionality and complexity [30]. One common approach for bending linear soft robots is derived from piecewise constant curvature modeling, based on which classical control strategies can be employed [31]. Despite the promising results, for more complex soft robots, the method is often ill-suited, as the dynamic behavior with the environment is not captured. Apart from these typical problems, overcoming the soft actuator's reliable extension under loading adds new challenges, such as buckling and undesirable bending. Given the gap in control strategies for complex soft robots, we used one of the common approaches of outsourcing some of the control to the robot's morphology and design [32].

In this paper, we embed the aforementioned requirements in a helically configurable soft pneumatic actuator that applies controlled multi-modal mechanostimulation. We introduce the



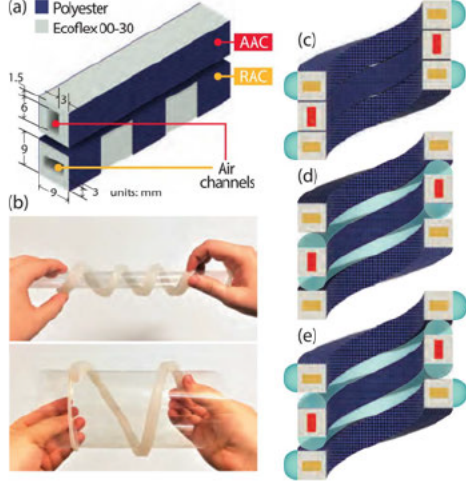


Fig. 2. SoPHIA Design. (a) Pneumatic chambers with embedded polyester that constraints inflation. (b) The pneumatic chambers can be configured into a helix around structures of varying diameter. (c) Cross-sectional area of SoPHIA when the RAC is inflated. (d) Cross-sectional area of SoPHIA when the AAC is inflated. (e) Cross-sectional area of SoPHIA when both chambers are inflated.

following contributions to this area of research: (1) introduction of the concept of coiling soft assembly for realizing deployable, multi-modal, compact, soft yet strong, and adaptable soft robotic implants; (2) modeling and mechanical characterization of a soft helical actuator for axial elongation and radial expansion to deliver mechanostimulation to tissue; (3) a physical *in vivo* simulator of the biological tissue's stiffness and growth to allow for the evaluation of the helical actuator's performance; (4) a model-based multi-stage control of the axial elongation of the helical actuator according to the tissue's physiological response.

### III. SOPHIA'S CONCEPTUAL DESIGN

We designed a soft actuator that is based on the anatomy of the GI tract. Most of the GI tract presents two types of muscle layers with fibers oriented circularly and longitudinally [33]. Therefore, a soft actuator for these organs should apply traction forces to both ends of the muscle layers in order to optimize the quality of the engineered tissue.

SoPHIA is an entirely soft actuator, made out of two identical elastomeric pneumatic chambers; the axial actuation chamber (AAC) and the radial actuation chamber (RAC) (Fig. 2(a)). Each chamber has a length of 48 cm, reaching 10.7 cm of height when the chambers are helically arranged and unpressurized. SoPHIA has a total weight of 95 grams. The chambers are made of Ecoflex 00-30 (Smooth-on Inc.) and can be configured into a helix of different diameters, depending on the tubular organ where it is placed (Fig. 2(b)). They are wrapped in polyester fabric, which in the AAC restricts radial expansion, while in the RAC restricts axial elongation (Fig. 2(c-e)). The AAC expands to displace adjacent chambers,

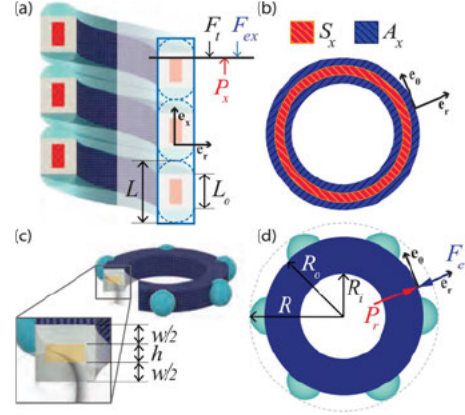


Fig. 3. Model of the simplified (a) side-view and (b) top-view axial elongation showing the interacting forces and areas respectively. (c) Perspective cross-sectional area and (d) top-view radial expansion showing the lengths and interacting forces respectively.

increasing the axial size of SoPHIA. The RAC exhibits laterally emerging balloons from the unconstrained sections, yielding radial expansion. These chambers are coiled together into a helical structure with interlayered actuation (Fig. 2(e)) (Fig. 1(b-c)). Details of the conceptual design, fabrication and Finite Element Modeling of the chambers are in [8]. Details on SoPHIA's fixation to *ex vivo* esophagus tissue are described in [7].

### IV. SOPHIA'S ANALYTICAL MODEL

In this section, an analytical model of the helical actuator is introduced and validated, to understand the relation between the physical components and their mechanical response to pressurization.

#### A. Analytical Model

For simplicity, the model is separated into two parts, one model for the AAC and one for the RAC. The main performance metrics are elongation and expansion. At equilibrium, the force balance equations projected respectively on  $e_x$  and  $e_r$  (Fig. 3) can be written as:

$$P_x S_x - F_{ex} - F_t = 0 \quad (1)$$

$$\int_{\theta_1}^{\theta_2} h P_r R d\theta - \int_{\theta_1}^{\theta_2} w \sigma_{er} R d\theta = 0 \quad (2)$$

where  $F_{ex}$  denotes the resistance force of the elastomeric walls to expansion along  $e_x$ ,  $F_t$  is the resistive force of the host tissue during the mechanotherapy treatment,  $P_x$  and  $P_r$  are the relative pressure of the air inside the air channels of the AAC and the RAC respectively.  $S_x$  is the cross-sectional area of the air channel of the AAC,  $h$  denotes the height of the air channel of the RAC,  $w$  is the thickness of the RAC walls and  $R$  is the external radius of the RAC,  $\sigma_{er}$  is the elastic stress of the elongated wall in the radial expansion of the robot,  $\theta_1$  and  $\theta_2$  are the angular limits of the area considered for the force



TABLE I  
MODEL PARAMETERS

| Notation | Description                                                            | Value | Unit            |
|----------|------------------------------------------------------------------------|-------|-----------------|
| $S_x$    | cross-sectional area of the inner channel of the AAC                   | 415   | mm <sup>2</sup> |
| $A_x$    | cross-sectional area of the elongated material in the AAC              | 829   | mm <sup>2</sup> |
| $w$      | width of the elongated material RAC                                    | 6     | mm              |
| $h$      | height of the RAC air channel                                          | 3     | mm              |
| $R_0$    | initial external radius of the RAC                                     | 2.49  | mm              |
| $R_i$    | internal radius of the RAC                                             | 17.5  | mm              |
| $L_0$    | initial height of the AAC                                              | 9     | mm              |
| $F_t$    | resistive force of the host tissue during the mechanotherapy treatment | 2     | N               |

balance equation, and finally  $d\theta$  is the angular differential for the integration.

We made the assumption that the two chambers can be decoupled and act independently in a single direction and that they deform uniformly. To simplify the analysis, we assumed that the coiled chambers behave as a stack of  $N$  circular chambers. The weight of the actuator and the weight of the air are neglected. Finally, we neglect any possible radial force applied by the tissue to the robot in order to match the characterization test presented in Section V.

Silicone rubbers are known as hyperelastic materials. Using a Neo-Hookean model for incompressible material in a uniaxial elongation to describe the elastic behavior of the chambers walls, equations (1) and (2), respectively, can be rewritten as follows:

$$A_x \frac{2E}{6} \left[ \left( \frac{L}{L_0} \right)^2 - \frac{L_0}{L} \right] + F_t = S_x P_x \quad (3)$$

$$w \frac{2E}{6} \left[ \left( \frac{R - R_i}{R_0 - R_i} \right)^2 - \frac{R_0 - R_i}{R - R_i} \right] = h P_r \quad (4)$$

$A_x$  is the cross-sectional area of the elongated material in the AAC,  $E$  is the Young's modulus of the material,  $L$  is the height of the AAC after elongation,  $L_0$  is the initial height of the AAC,  $w$  is the width of the elongated material in the RAC,  $R$ ,  $R_i$ , and  $R_0$  are respectively the external radius of the RAC once elongated, the internal radius of the RAC, and the initial external radius of the RAC. Fig. 3 illustrates the main parameters considered in the model.

The model parameters are given in Table I. In order to estimate the Young's Modulus, we used least squares curve fitting using the axial elongation results with the model obtained in equation (3). The Young's modulus was optimised in the range of values found in the literature, from  $E = 0.027$  MPa [34] to  $E = 0.069$  MPa [35]. After the fitting, the best Young's modulus was determined to be  $E = 0.068$  MPa. The values in Table I are based on the geometry of the actuator. The value for the resistance force  $F_t$  was taken from [12].

### B. Experimental Validation of the Model

Two experiments were run to validate the analytical model derived in the previous section. To validate the analytical

model for the AAC, we coiled three turns of that chamber around an oiled (Cole Parmer VacuumPump Oil CP 500) PLA tube to reduce friction and support it vertically without restricting its movement (Fig. 4(a)). An ABS plate at the top of the AAC acted as a reference for the two reflective distance sensors (GP2Y0A41SK0F, Sharp) to measure the elongation, as well as to support the weights that simulate the resistive force  $F_t$ . We used two distance sensors, averaging the readings, in order to avoid inaccuracies due to uneven elongation of the actuator on either side. We pressurized the AAC from 0 to 25 kPa and recorded the elongation readings to obtain  $L$ . We performed three trials. For the experiments, we define elongation as  $\frac{L-L_0}{L_0}$ .

To validate the RAC, we glued one turn of the RAC, along its entirely constrained face, around an ABS tube using cyanoacrylate adhesive (Fig. 4(c)). The ABS tube was supported by a clamp stand. Then, we pressurized the RAC from 0 to 20 kPa and measured its radial expansion, defined as  $\frac{R-R_0}{R_0}$  over three trials using ImageJ (NIH).

For both chambers, relative pressure was considered. The maximum pressure values were defined considering pressurization before anisotropic deformation and failure.

As can be seen in Fig. 4(b) and Fig. 4(d), there is a reasonable agreement between the theoretical models and the experimental values. The Root Mean Square Error (RMSE) between the three AAC elongation trials and the analytical model are 2.5%, 2.9%, and 3.6%. This means that on average, the error taken by using the model instead of the real data is 3.0 percentage points of elongation. The RMSE between the three RAC elongation trials and the model are 3.0%, 2.1%, and 3.2%; in this case the average error is 2.8 percentage points of expansion. The RMSE over trials shows that the error made per trial is consistent. The average error is similar for the two chambers but their ranges of expansion are different. Since the AAC has a longer stroke, the normalized standard error is smaller than for the RAC.

The observed differences, in both the AAC and the RAC, can be explained by the model limitations. First, the proposed model only takes into account the wall elongation in one direction during the pressurization, but does not consider the ballooning behavior of the chambers. Secondly, the Neo-Hookean model was chosen for its simplicity, in order to provide information about the impact of each design parameters, e.g., size, Young's modulus, on the actuator's elongation for a potential adaptation of SoPHIA to other applications.

## V. SOPHIA CHARACTERIZATION

We conducted experiments to characterize SoPHIA's elongation, expansion and force capabilities, to determine both its standalone capabilities, as well as those under conditions that are expected clinically. This section describes the corresponding experimental setups and methods. The electronic control platform is described in [8].

### A. Axial Elongation

Considering that SoPHIA's primary application is to elongate tissue via mechanostimulation, it is relevant to

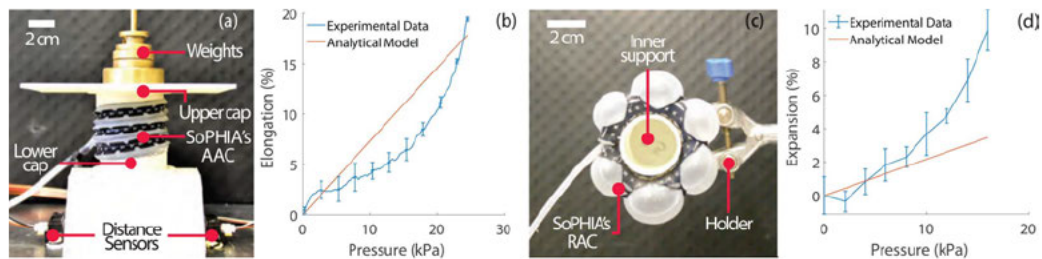


Fig. 4. Experimental validation of the mathematical model for SoPHIA's actuation chambers. (a) Side view of three AAC pressurized SoPHIA coils used to validate the mathematical model of axial inflation. (b) Experimental and analytical evaluation of the AAC actuation. (c) Top view of one of the RAC pressurized SoPHIA coils used to validate the mathematical model of radial inflation. (d) Experimental and analytical evaluation of the RAC actuation. Error bars stand for standard deviation of 3 trials. We pressurized the samples from 0 to 25 kPa for the AAC and from 0 to 22 kPa for the RAC. For the AAC, we used 2 N of load to represent the resistive force of the host tissue [12]. For the RAC, the resistive force was not considered.

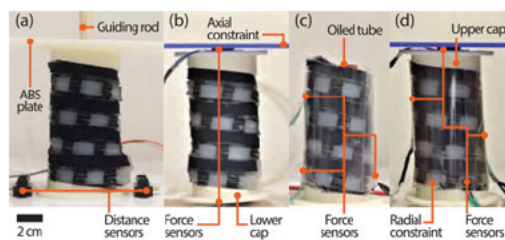


Fig. 5. Experimental setups to characterize elongation, expansion and force capabilities. (a) SoPHIA placed around an oiled tube and a rod that guides the axial free-load elongation of an ABS plate to record the actuator's elongation. (b) Two force sensors were placed at the top and bottom of an axially restricted SoPHIA to measure axial forces. (c) Three force sensors were placed on the expandable sections of the RAC to record forces exerted against a PET sheet that envelops SoPHIA. (d) SoPHIA restricted axially and radially to record overall forces.

characterize its axial elongation in dynamic conditions in order to evaluate its suitability to elongate tissue. In this set of experiments, we investigated the hysteresis of SoPHIA by measuring the free-load axial elongation response of the AAC, the RAC and the simultaneous pressurization (AAC+RAC) to pressures of 25 kPa, 20 kPa and a combined 55 kPa respectively. These were pressurization cycles of 2 seconds.

We placed SoPHIA around an oiled (Cole Parmer Vacuum Pump Oil CP 500) plastic tube to reduce friction and support it vertically without restricting its movement. A rod guided the axial elongation of an ABS plate that worked as a reference to the distance sensors to measure SoPHIA's elongation (Fig. 5(a)). This rod was necessary for the benchtop tests, yet not needed for the clinical setting when SoPHIA will be fixed to the tubular tissue. We performed three trials of five cycles each. Then, we inflated the AAC+RAC up to their breakage point to find their maximum expansion before failure, leaving time to settle. For this experiment, we only needed the oiled tube to support the actuator under free-load conditions (Fig. 5(a)). The elongation was recorded and then measured using ImageJ (NIH).

### B. Axial and Radial Output Forces

Given that SoPHIA may be used as an implantable device, it is necessary to understand the potential forces that it can exert against the target tissue. According to [12], forces to stretch the tissue axially should be around 2.3 N. In order to investigate if this force requirement is fulfilled, and considering the interdependence of the AAC and the RAC found in our previous work [8], we proceeded to characterize the force interactions between the two chambers under constrained and unconstrained setups, performing five trials for each set of conditions.

To measure the axial forces exerted by SoPHIA's elongation, we placed two force sensors on the top and bottom of the actuator respectively, and we restricted its expansion in these same two locations as shown in Fig. 5(b). Both force sensors were averaged to get an overall reading of the maximum axially exerted force. Then, we pressurized the AAC and the RAC independently and simultaneously. The constraints were placed in direct contact with SoPHIA. To measure the radial forces exerted by SoPHIA's expansion, the actuator was enveloped in a rigid polyethylene terephthalate (PET) cylinder with force sensors adhered to its inner surface and in direct contact with the unconstrained segments in the RAC, as shown in Fig. 5(c). Then, we pressurized the AAC and the RAC independently and then simultaneously. To measure the overall forces that SoPHIA can exert, the actuator was constrained both axially and radially at the same time (Fig. 5(d)). Then we pressurized the AAC and the RAC independently and then simultaneously.

### C. Structural Strength

Since SoPHIA needs to sustain forces over a long period of time, structural strength is an important feature to test, to determine how much weight the implant can support, at different inflation levels, without buckling. To test the structural strength, both the axial and radial chambers were inflated at 3 different levels of averaged pressure —10.5 kPa, 13.5 kPa and 16.5 kPa — as well as non-inflated. With no constraints around SoPHIA, weights were then progressively added, until we observed the actuator buckling. The weights were guided on to the top cap of SoPHIA by an acrylic tube. We repeated the experiment five times.



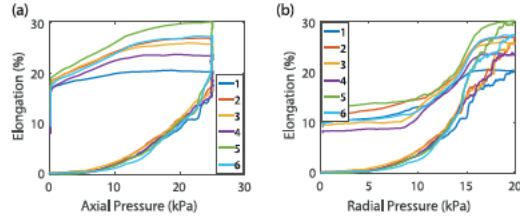


Fig. 6. SoPHIA's dynamic elongation behavior. (a) Elongation behavior as a function of axial pressure and (b) as a function of radial pressure when the AAC and the RAC are simultaneously actuated. SoPHIA exhibits hysteresis behavior and yields up to 30% of free-load axial elongation under pressurization cycles of 2 seconds. Experimental conditions are detailed in Section V-A.

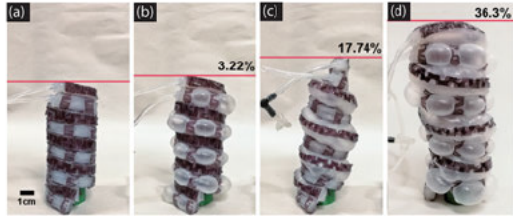


Fig. 7. SoPHIA's static behavior under different actuation modes. (a) SoPHIA in a relaxed state, (b) with only the RAC inflated, (c) only the AAC chamber inflated and (d) both chambers inflated. By inflating the AAC (25 kPa) and the RAC (20 kPa) simultaneously, SoPHIA is capable of reaching 36.3% of elongation in comparison to its original size.

## VI. RESULTS

### A. Axial Elongation

Under dynamic conditions, with two-second cycles, SoPHIA can axially elongate 0.25 cm at 25 kPa when only the AAC is pressurized, 0.7 cm at 20 kPa when only the RAC is inflated and 2.8 cm when both are pressurized simultaneously (Fig. 6). The elongation capability is enhanced by simultaneous pressurization of the chambers, exhibiting non-linear and hysteresis behavior. In comparison, under static conditions (when the actuator had time to settle) SoPHIA can axially elongate 3.2%, 17.7% and 36.3% of its original size when the RAC, the AAC and RAC+AAC are actuated respectively (Fig. 7).

### B. Axial and Radial Output Forces

SoPHIA is capable of exerting up to 7 N axially, when the AAC and the RAC are pressurized simultaneously at 38 kPa while entirely constrained (Fig. 8(c)), and it only exerts 0.69 N radially under the same conditions (Fig. 8(d)). This behavior is constant along all the setups, meaning that SoPHIA tends to direct its forces 90% more axially than radially. This is confirmed by looking at Fig. 8(a) and (b). This result is positive considering the intended clinical application as it means that SoPHIA is capable of exerting higher forces than the ones needed in proved functional mechanotherapy [12] at low levels of pressure. SoPHIA can exert up to 1.35 N of force at 38 kPa when it is radially constrained, which is equivalent to

the force we would be able to yield against a tubular organ (Fig. 8(b)).

### C. Structural Strength

SoPHIA's helical configuration provides increasing structural strength, with increasing levels of average pressure. In a relaxed state, SoPHIA can stand 500 g of weight without buckling (Fig. 9(a)). This strength increases as the actuator increases its inner pressure. Pressurized at an average of 16.5 kPa, it can withstand up to 800 g (Fig. 9(b)).

## VII. SOPHIA STAGED CONTROL

After characterizing SoPHIA, the control requirements for the real-life application were considered. Envisaging the *in vivo* tissue growth treatment, SoPHIA will be fixed at its ends inside the tubular organ and will inflate axially every 24 hours in order to apply tension to the tissue. During the week-long traction procedure, the physiological response of the tissue stiffening is to resist to the applied traction forces [16]. A release of this tension can be observed on the tissue between the traction applications, as a consequence of tissue relaxation and growth. In this section, we describe the setup, experimental procedure, results and SoPHIA's control used to simulate staged tissue growth mechanotherapy.

### A. Tissue Growth Simulation

1) *Physical Simulator*: A benchtop tissue growth simulator, shown in Fig. 10, was developed in order to derive the control of SoPHIA in an environment similar to the one *in vivo*. The benchtop simulator has two roles: to simulate the tissue stiffness and growth, and to monitor SoPHIA's performance. The tissue growth and stiffness are simulated via two mechanisms: a passive spring actuated plate, and a controlled lifting plate. Monitoring is then achieved using pressure, force and distance recordings from the platform.

Tissue growth and stiffness were simulated using a combination of passive and active mechanisms. The tissue stiffness is replicated using a compression spring (LP 026LM 06S316, Lee Springs) - which has a 0.05 N/mm spring rate, 50.8 mm free length, 7.035 mm solid length. As SoPHIA lengthens it experiences a resistant force corresponding to the level of spring compression. The spring itself was mounted around a plastic rod capped by two acrylic plates to prevent buckling.

The tissue growth was achieved using the active mechanism, made up of two NEMA-23 stepper motors (57STH56, Phidgets), controlled by two stepper motor drivers (TB6600, TopDirect) through an Arduino Uno R3 microcontroller. The two motors are coupled to two vertical M8 threaded bars which lift the moving plate. The moving plate itself carries the passive spring mechanism and, when displaced vertically, allows the spring to decompress, thus simulating tissue growth. The tissue growth is monitored via two distance sensors which measure the displacement of the moving plate.

Throughout the simulation, SoPHIA is also monitored by three different sensor readings: pressure, force and distance. The pressure recordings are taken from the dedicated AAC and RAC PCBs. Similarly to the previous experiments, the force

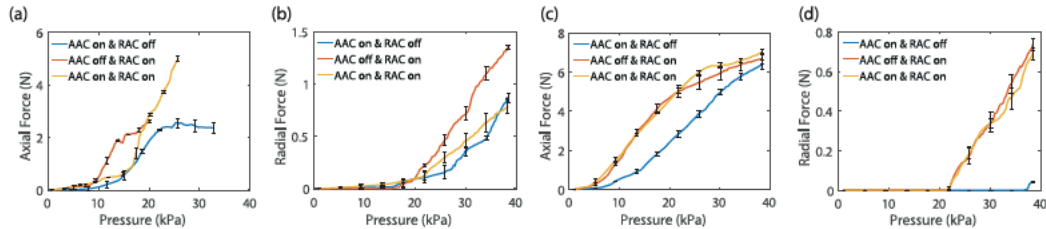


Fig. 8. Mean output force capabilities with standard deviation for five trials. (a) Axial forces under axial constraints and three actuation modes. (b) Radial forces under radial constraints and three actuation modes. (c) Axial forces under axial+radial constraints and three actuation modes. (d) Radial forces under axial+radial constraints and three actuation modes. As expressed in the results section, axial forces represent the most benefited from interdependence between chambers with the lowest standard deviation among trials and conditions. Experimental conditions are specified in Section V-B.

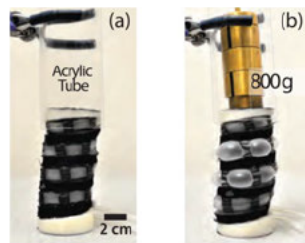


Fig. 9. SoPHIA's structural strength demonstrated (a) in a relaxed state and (b) at an averaged pressure of 16.5 kPa withstanding a maximum weight of 800 g.

recordings are taken from four force sensors, two placed on the top and two placed on the bottom. SoPHIA's elongation is measured by taking the average of two distance sensors which track the displacement between the spring actuated plate and the ground plate.

2) *Experimental Procedure:* We simulated the growth of 3.2 cm of virtual tissue by expanding SoPHIA axially, inflating simultaneously the AAC and the RAC. The elongation of both chambers was semi-constrained by the spring actuated plate, in order to simulate the resistance that SoPHIA will experience from the tissue under stretch. SoPHIA was constrained by a PET sheet, simulating its enclosure in a tubular tissue shown in Fig. 11(a). Both chambers were pressurized simultaneously, controlled by the target distance determined for each stage, shown in Fig. 11(a-e). The overall target distance of 3.2 cm was chosen as the maximal semi-constrained inflation elongation before risk of failure.

The procedure followed for the simulation of tissue growth consists of four stages described in Fig. 11. Firstly the actuator is inflated and then the moving plate is lifted to decompress the spring mechanism. Using this procedure, two sets of experiments were performed. The first set consisted of using manually tuned values for each stage that were then further improved in the second set using modeling and automatic tuning. Each set of experiments consisted of five trials. In this setup, SoPHIA exerts up to 0.19 N of force before decreasing as a result of the upwards displacement of the moving plate, simulating tissue relaxation at the end of each stage (Fig. 12). These results match with the previously stated experimental

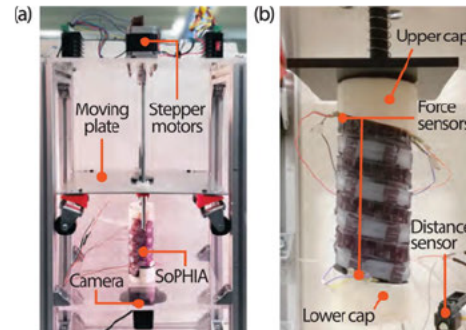


Fig. 10. Physical simulator of tissue stiffness and growth. (a) Overall side view; (b) detail of the sensors and actuator in the physical simulator.

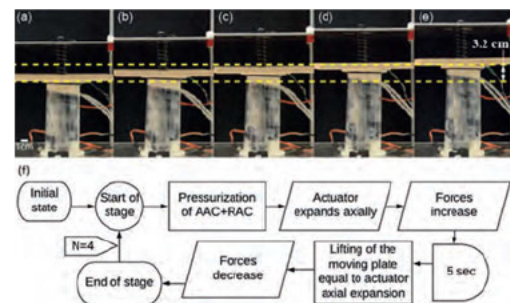


Fig. 11. Experimental procedure for the simulation of tissue growth treatment based on the staged application of traction forces. The dashed yellow lines represent the initial and final states of the moving plate. (a) SoPHIA in a relaxed state; (b) end of stage 1 reaching 0.8 cm, (c) end of stage 2 reaching 1.6 cm, (d) end of stage 3 reaching 2.4 cm, and (e) end of stage 4 reaching 3.2 cm. (f) Flowchart describing each stage in the simulation.

protocol in Fig. 11(a-e). The initial interaction force between SoPHIA and the tissue simulator was set to around 0 N, although could be tuned to other values depending on the initial position of the moving plate relative to SoPHIA.

### B. Modeling and Control for Axial Elongation

In this section, the control details of SoPHIA's axial elongation in the tissue growth simulator are explained. The



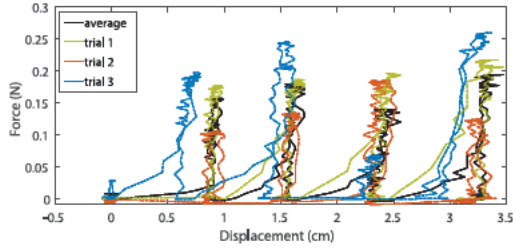


Fig. 12. Force response of SoPHIA caused by its elongation. The decrease in force represents the end of each of the stages during the simulation.

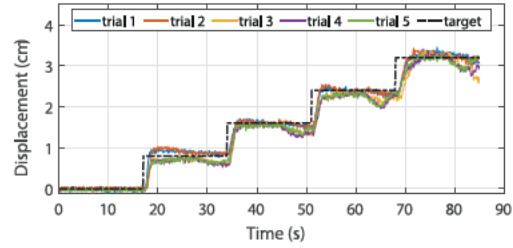


Fig. 13. Plot of all of the trials with the improved control throughout all 4 stages.

requirements of the controller are first established and more details about the controller used are given. The modeling approach is then presented, detailing how the transfer function models are derived. Finally, the control tuning is described and compared against the initial requirements.

1) *Specifications*: To establish the requirements for the controller design, both the clinical requirements and the current limitations of the actuator were taken into account. A steady-state error requirement of less than 1 mm was set. The steady-state error was limited to 1 mm, to be as small as possible, while taking into account the precision of the distance sensors that is around  $\pm 1$  mm. In addition, the rise time was set to be maximized to ensure a slow elongation of the tissue and to avoid tearing.

For the system itself, a position PI controller was used, as it was deemed appropriate to meet the requirements. A target elongation of 0.8 cm for each stage was chosen, to have 4 stages of visually noticeable elongation. Additionally, an overall safe pressure limit of 39 kPa was enforced, and stopped inflation if reached, to ensure SoPHIA would not burst. This overall safe pressure limit was determined from the maximum pressure of the previous experiments. Based on these design requirements and inspired by the current daily clinical intervention for tissue elongation [16], a gain-scheduled lengthening process was hypothesized to be suitable. Gain-scheduling was chosen to allow for better tracking of the reference signal and improve the overall behavior of the actuator. The PI values for all of the stages were initially set based on testing of previous actuators, before being improved later based on the modeling.

2) *Modeling*: For each of the stages, a model was found using system identification to aid with the tuning of the PI controller. Based on the linear dynamics observed in each stage, a low order linear time invariant transfer function was fit to the data. Of the five trials conducted, the best trial, selected based on minimal overshoots and minimal steady-state error for each stage, was chosen for modeling. A model was fit to this selected trial, identifying a second order model for stage 1 and first order models for the remaining stages. The transfer functions were determined using subspace identification with an automatic estimation of the initial condition and using the instrument variable approach for initialization.

3) *Control*: The PI values for each of the stages were tuned to meet the requirements set out in Section VII-B1. The overall

TABLE II  
AVERAGED PERFORMANCE CHARACTERISTICS FOR THE INITIAL AND IMPROVED TRIALS. TR: RISE TIME, OS: OVERSHOOT AND SSE: STEADY-STATE ERROR

|          | Initial |      |      |      | Final |      |      |      |
|----------|---------|------|------|------|-------|------|------|------|
|          | Stg1    | Stg2 | Stg3 | Stg4 | Stg1  | Stg2 | Stg3 | Stg4 |
| Tr (s)   | x       | 4.05 | 3.30 | x    | 1.12  | 1.20 | 1.90 | 2.45 |
| OS (%)   | 0       | 0.25 | 1.50 | 0.44 | 11.25 | 4.63 | 2.83 | 5.94 |
| SSE (cm) | 0.11    | 0.05 | 0.04 | 0.09 | 0.07  | 0.02 | 0.05 | 0.07 |

aim of the tuning process was to achieve a smaller steady-state error, while keeping rise time as slow as possible. The digital tuning of the PI values was done using the Ziegler-Nichols method applied to the developed model transfer functions. For the tuning, a small steady-state error was prioritized, as well as a slow rise time to achieve a steady inflation. After this initial digital tuning, the PI values were then further fine-tuned, taking into account the noticed behavior changes to smooth the response. Fig. 13 shows the comparison between initial and improved control throughout all 4 stages, where we demonstrate an improvement by implementing automatic tuning on the gain-scheduled lengthening process control scheme. The averaged achieved performance from the first five trials and the next five trials with improved control are summarized in Table II.

Overall, most of the stages showed a decrease in steady-state error or at least comparable values with the improved control, although this did come at the cost of greater overshoots and faster rise times. Additionally, compared to the initial tunings, where rise times were not available for stages 1 and 4 as 90% of the target value was not reached, all of the stages with the improved control reached these target values as a minimum. Overall, the system met the requirement of a steady-state error of less than 1 mm across all stages, promising safe operation for the envisaged clinical procedure.

## VIII. DISCUSSION & CONCLUSION

We introduced a Soft Pneumatic Helically-Interlayered Actuator (SoPHIA), capable of configurability and multimodal axial elongation and radial expansion, that we envisage as an implantable device to provide mechanostimulation for tissue growth. We evaluated SoPHIA's performance using a novel physical simulation platform that we envisage to be used as a research tool for tissue regeneration simulation.

We tested the capability of this actuator by evaluating the elongation, output force and characterizing its nonlinear behavior. SoPHIA is capable of growing axially 36.3% its original size under freeloading static conditions at an average pressure of 22.5 kPa and 33.3% under loaded-staged control at an average pressure of 21 kPa. By constraining SoPHIA entirely and then pressurizing both chambers, which is a clinically-realistic condition, SoPHIA can exert up to 7 N axially and up to 1.35 N radially, if only the RAC is pressurized. These capabilities make the actuator suitable to address the reconstruction of tissues in clinical conditions such as LGEA (where more than 3 cm of tissue is missing) or SBS.

While the output performance of the soft actuator is comparable to many soft pneumatic actuators [21], [36], the design and behavior of SoPHIA provide unique features that are clinically advantageous:

- By using a geometrically simple building block (actuation chambers) that can be coiled from one to a three-dimensional actuator, we can configure SoPHIA into a cylinder that fits to various diameters of hollow organs and tubular tissues, making it a versatile medical tool.
- Its ease of configuration could facilitate clinical intervention on different organs. This potentially could be done by performing a small incision near the target organ, through which the AAC and RAC could be inserted, implanted and removed employing minimally invasive surgery procedures such as laparoscopy.
- By varying the orientation of the chambers, we can address different conditions. For example, the RAC could be oriented to expand inwards in order to stimulate the intestine during SBS treatment. It can also be oriented to expand outwards during LGEA treatment (Fig. 1).
- The helical shape also provides structural support, with air pressure distributed uniformly across the diameter and height of the actuator. We showed that the implant can support up to 800 g without buckling. This feature is clinically important as buckling could lead to a misshaped organ.

Over the duration of the treatment, the AAC may be progressively inflated to support the lengthened tissue. The RAC may be intermittently inflated to provide radial stimulation to the tubular organ and also decrease fibrotic response. Powering and monitoring a robotic implant via loose cabling that exits the body through a skin port into a control box was demonstrated clinically viable in [12]. For the pneumatic tubing a similar approach is envisaged, however this needs to be tested pre-clinically. As presented in our previous work [7], adaptors can be used to attach SoPHIA without being in direct contact with the organ to heal, in order to reduce fibrosis. SoPHIA's helical configuration provides multi-modal behavior achieved by interlayering elastomeric chambers of different functions into the helix. Typically, soft helical actuators show some degree of torsion when they incorporate a backbone into their design [37]. SoPHIA is a zero-torsion helical actuator due to it being constrained axially and radially.

From the dynamic elongation tests, we observed that the RAC generates much more elongation than the AAC, whilst

under static conditions the opposite is true. We hypothesize that this difference comes from the AAC having a much slower response time, partly caused by the stacking of coils. As the RAC showed superior dynamic expansion than the AAC, which is desirable for mechanotherapy, one option is to use the RAC design for axial inflation as well. Another alternative to increase SoPHIA's axial elongation is to decrease shear stress between the chambers caused by their different expansion rates that currently causes the actuator to burst.

We evaluated the behavior of SoPHIA using a novel physical simulator. During a week-long clinical tissue lengthening treatment, the tissue may become stiffer (due to inflammation), relax (due to its viscoelastic properties) or grow. These mechanical and physiological changes are being emulated by the physical simulator. Such simulations are difficult to achieve with current phantom tissues or with *ex vivo* biological tissues, as they exhibit limited lengthening. This makes the physical tissue growth simulator ideal to dynamically represent different states of tissues, from mechanical to metabolic, under either physiological or pathological conditions. Further work still needs to be done to simulate the radial growth, as well as the real-time stiffness variation of the tissue.

While the soft-matter actuator is desirable for its stretchability to support a growing tissue, its intrinsic nonlinear behavior needs to be precisely controlled for clinical safety. We thus developed a staged-position gain-scheduled controller, which emulates the daily tissue stretch within the existing clinical treatment. We successfully evaluated the implant control in a physiologically-relevant scenario. Variability of the system represents an important challenge when controlling it due to the sensor disturbances and elastomer non-linearity and wear. However, the presented analytical model constitutes a reliable tool for the prediction of the AAC and the RAC behavior, as there is a reasonable agreement between the theoretical models and the experimental values (Fig. 4(b) and Fig. 4(d)). Although the proposed model does not consider the ballooning behavior of the chambers, it provides qualitative information of the design parameters that impact SoPHIA's elongation. In the future other nonlinear models, such as Ogden, will be investigated in order to provide more detailed performance insights.

Due to the lack of kinematic equations that describe soft robots in general [31], our approach was a combination of using traditional control techniques and designing constraints to achieve the desired actuation. However, further research is still needed in order to model the behavior of soft machines from first principles. Consequently, we may be able to tune the controller better to decrease the steady-state error, while keeping large rise times. Force control could also be used, such that SoPHIA applies constant traction forces and adapts to the tissue response.

Future directions include embedding soft sensors to record SoPHIA's elongation and forces, in order to carry out more accurate mechanotherapy control *in vivo*. Further work also includes the advancement of the axial elongation capability of SoPHIA for a larger capacity of tissue lengthening.

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