Bio-Inspired Soft Artificial Muscles for Robotic and Healthcare Applications

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Phan, Phuoc Thien

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Bio-Inspired Soft Artificial Muscles
for Robotic and Healthcare Applications

Phuoc Thien Phan

A thesis submitted to fulfil the requirements for the conferral of the degree of Doctor of Philosophy. This research has been conducted with the support of the University International Postgraduate Award (UIPA), UNSW Sydney.

August 2023
Declarations

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<td>Twisting and braiding fluid-driven soft artificial muscle fibers for robotic applications</td>
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<tr>
<td><strong>Authors:</strong></td>
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<tr>
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I confirm that where I have used a publication in lieu of a chapter, the listed publication(s) above meet(s) the requirements to be included in the thesis. I also declare that I have complied with the Thesis Examination Procedure.
Acknowledgements

The past four years have been an unforgettable journey for me. I learned new things, met amazing people, enjoyed wonderful nature and experienced Australia’s inclusive culture. I believe the journey has broadened my horizons and made me a better person. I am very grateful and honored to receive financial support from the University of New South Wales, which has given me the opportunity to pursue my academic dream in Australia.

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Abstract

Soft robotics and soft artificial muscles have emerged as prolific research areas and have gained substantial traction over the last two decades. There is a large paradigm shift of research interests in soft artificial muscles for robotic and medical applications due to their soft, flexible and compliant characteristics compared to rigid actuators. Soft artificial muscles provide safe human-machine interaction, thus promoting their implementation in medical fields such as wearable assistive devices, haptic devices, soft surgical instruments and cardiac compression devices. Depending on the structure and material composition, soft artificial muscles can be controlled with various excitation sources, including electricity, magnetic fields, temperature and pressure.

Pressure-driven artificial muscles are among the most popular soft actuators due to their fast response, high exertion force and energy efficiency. Although significant progress has been made, challenges remain for a new type of artificial muscle that is easy to manufacture, flexible, multifunctional and has a high length-to-diameter ratio. Inspired by human muscles, this thesis proposes a soft, scalable, flexible, multifunctional, responsive, and high aspect ratio hydraulic filament artificial muscle (HFAM) for robotic and medical applications. The HFAM consists of a silicone tube inserted inside a coil spring, which expands longitudinally when receiving positive hydraulic pressure. This simple fabrication method enables low-cost and mass production of a wide range of product sizes and materials. This thesis investigates the characteristics of the proposed HFAM and two implementations, as a wearable soft robotic glove to aid in grasping objects, and as a smart surgical suture for perforation closure. Multiple HFAMs are also combined by twisting and braiding techniques to enhance their performance.

In addition, smart textiles are created from HFAMs using traditional knitting and weaving techniques for shape-programmable structures, shape-morphing soft robots and smart compression devices for massage therapy. Finally, a proof-of-concept robotic cardiac compression device is developed by arranging HFAMs in a special configuration to assist in heart failure treatment.

Overall this fundamental work contributes to the development of soft artificial muscle technologies and paves the way for future comprehensive studies to develop HFAMs for specific medical and robotic requirements.
Publications

*Publications used in this thesis

Patents


Journal articles


Conference proceedings


Media coverage

[M5] **Flexible 3D bioprinter, 2023**


[M4] **Smart textiles, 2022**


[M3] **Smart surgical sutures, 2021**

Featured on Engineers Australia, AZO Materials

[M2] **Helical soft fabric gripper, 2020**

Featured on UNSW Newsroom, IEEE Spectrum, ASME, Engineers Australia, ACM TechNews, TechXplore, Scienmag, ScienceDaily, Futurism, CNET, New Atlas,
Advanced Science News, EurekAlert!, Gizmodo, Mirage News, Medianet, Big Think, Scimex, Biomimicry Institute, Cosmos, Nanowerk, Noticias de la Ciencia, ITmedia NEWS, 7thSpace, Sott.net, North Queensland Register, Queensland Country Life, The Land, Farm Weekly, Ranger Rick Magazine.

[M1] **Wearable haptic device, 2020**

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<tbody>
<tr>
<td>ACE</td>
<td>angiotensin-converting enzyme</td>
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<tr>
<td>AMF</td>
<td>artificial muscle fiber or artificial muscle filament</td>
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<td>APC</td>
<td>artificial pericardium</td>
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<td>BiVAD</td>
<td>biventricular assist device</td>
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<tr>
<td>BLE</td>
<td>Bluetooth low energy</td>
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<tr>
<td>bpm</td>
<td>beats per minute</td>
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<td>B-SMAM</td>
<td>braiding configuration of soft microtubule artificial muscle</td>
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<td>CABG</td>
<td>coronary artery bypass grafting</td>
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<td>CAD</td>
<td>coronary artery disease</td>
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<td>CNT</td>
<td>carbon nanotube</td>
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<td>CSD</td>
<td>cardiac support device</td>
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<td>DCCD</td>
<td>direct cardiac compression device</td>
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<td>DEA</td>
<td>dielectric elastomer actuator</td>
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<td>DNA</td>
<td>deoxyribonucleic acid</td>
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<tr>
<td>DOF</td>
<td>degree-of-freedom</td>
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<td>EAP</td>
<td>electroactive polymer</td>
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<td>ECG</td>
<td>electrocardiogram</td>
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<td>EFTR</td>
<td>endoscopic full-thickness resection</td>
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<td>EGaIn</td>
<td>eutectic gallium-indium</td>
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<td>EMVA</td>
<td>electromagnetic vibrotactile actuator</td>
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<td>EPE</td>
<td>elastic potential energy</td>
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<td>endoscopic submucosal dissection</td>
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<td>endoscopic sleeve gastroplasty</td>
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<tr>
<td>FEA</td>
<td>finite element analysis</td>
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<td>fluidic fabric muscle sheet</td>
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<td>GI</td>
<td>gastrointestinal</td>
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<td>GUI</td>
<td>graphical user interface</td>
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<td>heart failure</td>
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<td>hybrid yarn artificial muscle</td>
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<td>IC</td>
<td>integrated circuit</td>
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<td>IDE</td>
<td>integrated development environment</td>
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<td>IOP</td>
<td>incisionless operating platform</td>
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<td>IPAM</td>
<td>inverse pneumatic artificial muscle</td>
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<td>ionic polymer-metal composite</td>
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<td>LCP</td>
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<td>LMMS</td>
<td>liquid metal microtubule sensor</td>
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<td>LV</td>
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<td>natural orifice transluminal endoscopic surgery</td>
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<td>over-the-scope clip</td>
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P2PE  peak-to-peak error
PAM  pneumatic artificial muscle
PCB  printed circuit board
PCD  portable compression device
PDMS  poly-dimethylsiloxane
PET  polyethylene terephthalate
PLA  polylactic acid
PLM  pressure locking mechanism
POSE  primary obesity surgery endoluminal
POTS  postural orthostatic tachycardia syndrome
PSO  particle swarm optimization
PTFE  polytetrafluoroethylene
PVCD  passive ventricular constraint device
PVDF  polyvinylidene fluoride
RCCD  robotic cardiac compression device
RMSE  root mean square error
RV  right ventricle
RVAD  right ventricular assist device
SAM  soft artificial muscle
SMA  shape memory alloy
SMAM  soft microtubule artificial muscle
SMP  shape memory polymer
STAM  soft tendon-like artificial muscle
S²  smart surgical
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<th>Description</th>
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<tr>
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<td>straight configuration of soft microtubule artificial muscle</td>
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<tr>
<td>T-SMAM</td>
<td>twisting configuration of soft microtubule artificial muscle</td>
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<tr>
<td>TAH</td>
<td>total artificial heart</td>
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<tr>
<td>VAD</td>
<td>ventricular assist device</td>
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<tr>
<td>VAMP</td>
<td>vacuum-actuated muscle-inspired pneumatic</td>
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<td>VTE</td>
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Chapter 1

Introduction

1.1. Research overview

Research interest in soft robotics, especially soft artificial muscles (SAMs) for robotic and medical applications, is growing rapidly. SAMs are soft actuators made of soft, flexible, deformable and compliant materials. They exhibit excellent abilities to cope with changing environments, are resilient to disturbance and are safe for human interaction, allowing them to potentially replace their rigid counterparts in many robotic and medical applications. SAMs, depending on their composition, can be controlled by a variety of stimuli, such as electricity, magnetic fields, temperature and pressure. Current SAM technologies have a lot of room for improvement. For example, electrically-driven SAMs based on phase-change materials and thermally-driven SAMs have the limitations of slow response time and low energy efficiency. Dielectric elastomer actuators require high voltage to operate. Magnetically-driven SAMs have weak actuation forces and bulky external driving sources. Although pressure-driven SAMs are known for their high force exertion and high energy efficiency, most of them have limited strain and require bulky and noisy pneumatic pumps to operate. There is a need for SAMs that are easy to fabricate and conform to similar performance characteristics as human muscles. However, the development of SAMs that are flexible, multifunctional and have a high length-to-diameter ratio remains challenging.

Inspired by human muscles, this research proposes a soft, scalable, flexible, multifunctional, responsive, and high aspect ratio hydraulic filament artificial muscle (HFAM), which can be programmed for both motion and force. The HFAM is created by inserting a miniature silicone tube inside a helical coil spring, which connects to a fluid transmission tube to receive fluid pressure. HFAM, once powered by hydraulic pressure, can produce high elongation and high energy conversion efficiency. Its elongating and contracting movements are beneficial for manipulating links and joints. The HFAM can be used as a monofilament soft actuator for multi-purpose actuation mechanisms or combined by twisting and braiding to enhance its
performance and open up new possibilities. Furthermore, its monofilament form can also be exploited to create smart textiles using conventional knitting and weaving techniques for a broad spectrum of robotic and medical purposes.

The proposed HFAM is implemented in a wide range of robotic and medical applications. Specifically, HFAM is used to actuate a wearable soft robotic glove to assist in grasping objects. It is also utilised to develop smart surgical sutures for perforation closure. Smart textiles made from HFAMs have the ability to change shape to create bio-inspired shape-morphing soft robots. HFAM-based smart woven sheets can also be implemented into smart compression devices for compression therapy. A special configuration of the HFAM is used to construct a robotic cardiac compression device to aid in the treatment of heart failure. This wide range of applications demonstrates the capabilities and versatility of HFAM. This research serves as a fundamental investigation that paves the way for further comprehensive studies to develop HFAMs for specific medical and robotic requirements.

1.2. Research aims

The development of HFAMs and their versatile configurations for robotic and medical applications can be divided into the following aims.

- **Aim 1**: Design, manufacture, examine the characteristics, develop mathematical models and evaluate the performance of the proposed HFAM.

- **Aim 2**: Demonstrate the potential applications of HFAM in its monofilament form through a wearable soft robotic glove and smart surgical sutures.

- **Aim 3**: Investigate HFAM’s twisted and braided configurations to enhance performance and expand their capabilities.

- **Aim 4**: Create HFAM-based smart textiles using knitting, weaving and sticking techniques, and demonstrate the concept of smart compression devices, programmable shape-shifting structures and bio-inspired shape-morphing soft robots.

- **Aim 5**: Develop a nonlinear hysteresis model and control scheme for woven smart textiles.

- **Aim 6**: Develop a proof-of-concept robotic cardiac compression device made of HFAMs to assist in the treatment of heart failure.
1.3. Thesis structure

The following thesis structure closely matches the development of the proposed HFAMs, their configurations and their implementation in robotic and medical applications. An overview of the thesis structure is shown in Fig. 1.1 and outlined below.

Chapter 1 introduces the research topic, objectives and structure of the thesis.

Chapter 2 provides a detailed literature review of soft artificial muscle technologies and four targeted application areas including wearable assistive devices, wound closure technologies, shape-morphing soft robots and smart textiles, and cardiac assist devices.

Chapter 3 investigates the design, fabrication, experimental characterisation, and analytical modeling of the proposed HFAM. This chapter also demonstrates the use of HFAMs to actuate a wearable soft robotic glove.

Chapter 4 presents the application of HFAMs in smart surgical sutures for some common surgical procedures such as perforation closure and tissue folding.

Chapter 5 studies the twisting and braiding configurations of HFAMs to enhance elongation and force performance. This chapter also demonstrates a tubular muscle created by the round braid technique for various applications.

Chapter 6 presents the ability of HFAMs to construct smart textiles using knitting, weaving and sticking techniques. The shape-changing ability of smart textiles is also demonstrated.

Chapter 7 introduces a nonlinear hysteresis model, a feedforward controller and an adaptive controller to track the output elongation of a woven smart textile. This chapter also demonstrates a portable smart compression device for compression therapy to improve blood circulation.

Chapter 8 introduces a proof-of-concept robotic cardiac compression device made of HFAMs to assist in the treatment of heart failure. The proposed device can be integrated with an artificial pericardium for additional functionality.

Chapter 9 discusses the research results, limitations and future directions.
Figure 1.1. Overview of the thesis structure corresponding to the development of hydraulic filament artificial muscles and their versatile configurations.
Robotic technologies have become increasingly prevalent in the healthcare system worldwide. In particular, surgical robots have revolutionised operating rooms with the ability to provide safer surgical procedures and reduce the risk of complications, thus bringing enormous benefits to patients [1-4]. Another example is robot-assisted physiotherapy to rehabilitate people who have had a stroke or spinal cord injury [5-8]. The fundamental actuation of these robotic systems is a tendon-driven mechanism known for its simple implementation and high-force transmission [9]. However, it has the disadvantage of high friction loss and backlash hysteresis, which causes complexity of the driving system and control strategy [10].

Soft robotics and soft actuator technologies have the potential to create a breakthrough actuation mechanism to enhance the performance of the current tendon-driven actuation and open up new possibilities for improving the medical and healthcare system [11]. Research interest in soft robotics has grown exponentially over the last two decades, resulting in a great number of impressive discoveries in soft actuators, soft surgical robots, wearable devices, soft grippers and soft sensors [12, 13]. Challenges remain in many aspects of soft actuators, including limited force generation, difficulty sensing and monitoring, the complexity of precise and reliable control and limited durability. A miniature, long, flexible and multifunctional soft actuator is required for a wide range of robotic and medical applications. However, as research and development continue, it is expected that more impactful soft robotic technologies will emerge in the future.

This literature review presents the state-of-the-art soft actuator technologies in four main categories based on their external driving sources including electrically-driven, magnetically-driven, thermally-driven, and pressure-driven actuation. This chapter also introduces an overview of four potential robotic and medical fields targeted for the implementation of the proposed soft actuator. These include wearable assistive devices for rehabilitation and human
argumentation, smart surgical sutures for perforation closure, smart textiles for compression devices and shape-morphing soft robots and assistive devices for heart failure (HF).

2.1. Soft artificial muscles

Soft robotics is an emerging research area that has attracted considerable attention from researchers, scientists and dedicated journals worldwide [11, 14, 15]. This interdisciplinary field draws on expertise from mechatronics and biomedical engineering, material science and computer science to create soft robots made of flexible and compliant materials, enabling them to interact safely and effectively with humans and their environments. The soft and flexible characteristics of soft robots make them ideal for applications where conventional rigid robots are unsuitable, such as wearable devices, medical devices and search and rescue operations [16, 17].

Soft actuators or soft artificial muscles (SAMs) are key components of soft robotics and are used to generate the motion and force of soft robots. Recently, SAMs have arisen as an excellent candidate to potentially replace rigid actuators in many robotic and medical applications due to their flexibility, adaptability to dynamic environments, resilience to disturbances and human-friendly interaction [18-20]. These compliant properties of SAMs can be attributed to material advancement and structural design. The former factor has been demonstrated by the introduction of a vast number of amazing soft materials such as shape memory polymers (SMPs), liquid crystal polymers (LCPs), electroactive polymers (EAPs), hydrogels, silicone elastomers and textile-based materials [19, 21, 22]. Besides being soft and flexible, these materials are also responsive to stimuli, meaning they can change shape or size in response to a specific excitation. Related to structural design, researchers focus on geometric arrangements and shape configurations of features or components of SAMs to generate the desired movement, deformation and force [18, 23].

SAMs offer new possibilities for flexible and adaptable technology and have the potential to revolutionise many research and industrial areas through a wide range of potential applications. For example, SAMs can be used to create flexible and adaptable soft robots such as soft manipulators [24-26], locomotion systems [27] and soft grippers [28] to work in unstructured environments or arbitrarily shape objects while providing safe human-machine interaction. In the medical field, SAMs can be implemented into wearable assistive devices for people with
disabilities or human augmentation and soft surgical instruments for safer surgical procedures. In addition, smart garments made of SAMs can provide therapeutic compression or adjust to the wearer’s movements for use in sports, healthcare and military industries [29-31].

External excitations or stimuli to operate active materials play an essential role in soft robotic systems. The choice of external stimuli depends on applications and the required characteristics of SAMs. The combination of a SAM and its excitation source determines the characteristics and performance in terms of strain, stress, hysteresis and responsiveness of a soft robotic system [19, 32, 33]. There are broadly four primary sources of excitation, including electricity, magnetic fields, temperature and pressure, that have been widely studied and employed in the field of soft robotics (Fig. 2.1). A review of these stimuli-responsive SAMs is presented in the next section. Other types of actuation, such as chemical and light have also been exploited in some studies. However, their limitations of low force exertion and slow response time hinder their use in practical applications. More details can be found in reference [19].

2.1.1. Electrically-driven SAMs

Electrically-driven SAMs rely on the responsive characteristic of soft materials to electrical stimulation. When receiving electrical power, these materials change shape and size, generating motion and force [34]. This type of actuation has been greatly developed due to the versatile and easy modulation of electric signals as well as the rapid development of electrically-responsive materials. Specifically, electroactive polymers (EAPs) are perhaps the most popular electrically-responsive materials. They are very soft, flexible and compliant materials that can induce mechanical motion in response to an electric field. EAPs can provide high energy efficiency and moderate strains [24].

There are generally three types of EAPs: ionic, dielectric and conductive. Ionic polymer-metal composites (IPMCs) are made of a thin and flexible film coated with metal electrodes on both sides. When an electric field is applied to the metal electrodes, ion exchange occurs between the electrodes and the surrounding electrolyte, causing the film to swell, which results in a bending or twisting motion. IPMCs can provide high strain and require low driving voltages but have highly unstable deformation properties, low durability and mechanical strength [24, 35]. Dielectric elastomer actuators (DEAs) consist of a thin and compressible membrane sandwiched between two electrodes. Under high voltage, the electrostatic interaction of the two electrodes deforms the membrane, creating motion (Fig. 2.1A). DEAs offer versatile performance through the customisation of membrane materials and thicknesses. DEAs are
responsive and can produce 100% strain [36, 37]. However, they require a high operating voltage (several kilovolts), which can pose safety concerns for human use. Conductive polymers, which won the 2000 Nobel Prize in Chemistry [38], are biocompatible and have low operating voltages. They can produce moderate force but have slow response times and need to be immersed in an electrolyte [39]. Bay et al. reported a 50 mm conductive polymer strip that achieved 8% strain in 20 s when receiving 1.5 V [40].

In a recent approach, electrically-driven SAMs made of phase-change materials require low operating voltage but can generate high strain of up to 900% [41] and high force of 4.5 kN [42]. The heat generated by electric current causes elastic materials to change their phase (due to the vapor of the encapsulated liquid), resulting in a large volume change that induces motion and force (Fig. 2.1B). However, these SAMs have the disadvantage of low energy efficiency and slow response times [41, 43].

2.1.2. Magnetically-driven SAMs

Magnetically-driven SAMs are typically made of a composite of magnetic particles and soft, uncured silicone elastomers. Upon exposure to an external magnetic field (permanent magnet or electromagnet), the alignment movement of the embedded magnetic particles generates motion and force [44, 45]. Depending on the distribution and magnetisation of these particles, which can be programmed at the fabrication stage, different desired motions and deformations of the actuators can be achieved. The base materials are soft and flexible polymers such as poly-dimethylsiloxane (PDMS), Ecoflex and polyurethane.

Lum et al. presented a jellyfish-like robot with two soft, programmable tentacles made from a soft composite material [44]. The tentacles could be manipulated to bend back and forth by an external magnetic field. Diller et al. developed a miniature swimming robot from permanent magnetic particles and Ecoflex that could be actuated by an external rotating magnetic field (Fig. 2.1C) [46]. Do et al. introduced soft electromagnetic vibrotactile actuators (EMVAs) and a miniature soft robotic gripper made from a low melting point alloy, permanent magnets, magnetic microparticles and a soft silicone substrate for manipulation with various objects (Fig. 2.1D) [45]. Hu et al. developed a small-scale soft robot that could perform multimodal locomotion, including swimming, climbing, rolling, walking, jumping and crawling using a programable external magnetic field [47].

Thanks to the wireless magnetic interaction, magnetically-driven SAMs are useful for many applications in confined spaces or requiring locomotion such as microsurgery and drug
delivery. This type of SAM is responsive but has the limitation of weak actuation force and bulky external driving source. The magnetic force decreases significantly as the distance between the actuators and the external source increases. In addition, their operation is greatly affected by the surrounding environment, especially in magnetic environments or nearby ferromagnetic materials [48].

2.1.3. Thermally-driven SAMs

Thermally-driven SAMs are made of heat-responsive materials whose deformation can be controlled by an external heat source such as resistive heaters, visible and infrared light [49]. The thermal conductivity or thermal response characteristics of SAM materials determine the actuation speed and generated strain. Thermally-driven SAMs can be activated remotely, allowing for the simplification of their structure and the possibility of miniaturisation of the actuator. Jiang et al. introduced a bilayer robot (1 × 7 mm) made from PDMS and graphene nanoplatelets that could be deflected by 1.5 mm for multiple cycles using near-infrared light within 3.4 s [50]. Li et al. reported the use of flexible conductive nanocomposite polymer heaters to induce volume changes in hydrogels to manipulate a flexible microvalve diaphragm actuator [51]. Na et al. used thermally-responsive hydrogels as an active layer of a trilayer-hinge to fabricate self-folding origami (Fig. 2.1F) [52].

Shape memory polymers (SMPs) and shape memory alloys (SMAs) are a class of materials that tend to restore their pre-programmed shapes when heated above their transition temperatures. SMPs and SMAs have shown excellent performance with high elongation of up to 1000% [53] and high stress of 200 MPa [54] at the trade-off with slow actuation speed, low energy efficiency and high hysteresis. In another approach, Haines et al. created a new type of temperature-actuated SAM by twisting fishing lines and sewing threads. When heated with hot water, it could shrink 49% of its original length at 1 MPa load and exert a stress of 140 MPa at 4.5% contraction (Fig. 2.1E) [26]. Despite its high strength, this type of SAM has the drawbacks of slow response time and low energy efficiency [51, 52].

2.1.4. Pressure-driven SAMs

Pressure-driven SAMs are probably the most popular used soft actuators. They generally have a simple structure and are easy to fabricate using low-cost manufacturing methods. Nevertheless, they exhibit impressive performance with high strain, high force generation and high energy efficiency [19, 20, 33, 55, 56]. Pressure-driven SAMs typically consist of a soft, deformable and hollow body, a pressure transmission medium and a pressure source. When
pressure is applied to the body, the SAMs generate movements depending on their structural design. A stiffer wall can resist bending better than a softer one, resulting in a bending motion towards the stiffer layer. In other aspects, SAMs can elongate, expand and contract upon pressurisation. The medium of pressure transmission can be pneumatic or hydraulic. Light and compressible air or gas can reduce response time and produce loud noises, whereas incompressible hydraulic fluid can produce high force with high-frequency responses.

The McKibben pneumatic artificial muscles (PAMs) are arguably the most widely recognised SAM within this pressure-driven class. These soft actuators consist of a mesh structure resembling a sleeve enclosing an elastic hollow tubing [57-59]. Upon application of a positive pressure source, the McKibben PAMs undergo radial expansion and axial contraction simultaneously, generating a contraction force along its axis (Fig. 2.1H). In a study conducted by Kurumaya et al., it was shown that a slender McKibben muscle (outer diameter 1.8 mm) could generate a contraction force of 9.3 N and a contraction ratio of 21% at supplied air pressure of 0.3 MPa [59]. By braiding several individual McKibben PAMs together, the contraction ratio and force can be further improved [59-61].

Belding et al. introduced a new fluid-driven SAM by modifying the McKibben muscles, in which the outer mesh was replaced by a plastic tube with pre-cut splits [62]. These splits are strategically arranged so that the soft actuator can provide desired movements such as bend, twist, contract and sequential actuation. Hawkes et al. created an inverse pneumatic artificial muscle (IPAM), consisting of an elastic tube surrounded by a layer of non-stretchable filaments, which exhibited over 300% strain under pressurisation and 419 kPa stress when the pressure was released [63]. Zhu et al. introduced a fluidic fabric muscle sheet (FFMS) created by inserting a silicone elastomer tubing inside a pre-made fabric channel. The channel geometry is based on applications and can be easily accomplished using computerised sewing techniques. The FFMS could produce multimodal motion including stretching, bending, contracting and conforming to arbitrarily shaped objects [64]. Besides positive pressure, vacuum power has also been explored to induce contraction motion and force of SAMs. Li et al. developed fluid-driven origami-inspired artificial muscles (FOAMs) from a compressible skeleton encapsulated in flexible sheets which, powered by vacuum pressure, provide 90% strain, 600 kPa stress and 2 kW/kg power density [65]. Yang et al. created vacuum-actuated muscle-inspired pneumatic structures (VAMPs) from elastomeric materials that are capable of generating linear motion through a combination of vacuum pressure and mechanical buckling (Fig. 2.1G) [66]. The study showed the VAMPs could reach 45% strain and 65 kPa stress.
Figure 2.1. Four types of soft artificial muscles (SAMs) based on their driving sources. (A) Dielectric elastomer actuators [19]. (B) SAM made of phase-change composite [41]. (C) Magnetically-driven swimming robot [46]. (D) Soft robotic gripper made of soft electromagnetic vibrotactile actuators [45]. (E) Hydrothermal lifting of a coiled fishing line [26]. (F) Thermally-driven self-folding origami [52]. (G) Vacuum-actuated muscle-inspired pneumatic structures [66]. (H) McKibben pneumatic artificial muscle [58].
2.2. Robot-assisted devices for rehabilitation

2.2.1. Clinical need

Tendon-driven or cable-driven actuation has been recognised as an important actuation mechanism in many robot-assisted surgical systems such as the Da Vinci surgical system [67] and the Medrobotics Flex system [68], where conventional rigid actuators seem unsuitable because of their bulky size and heavy weight [2]. Similarly, tendon-driven mechanisms are widely adopted in the field of wearable assistive devices due to their power, flexibility and compactness [69, 70]. The tendon-driven mechanism transmits the required force and position from external actuators to remote joints utilising either pulleys or flexible sheaths. Many studies have shown that the tendon-sheath mechanism exhibits a superior ability to handle unpredictable bends in the human gastrointestinal (GI) tract, making it widely used in surgical robotic applications and flexible surgical instruments [71]. However, the nonlinear friction and backlash hysteresis between the sliding tendon and routing sheath can lead to significant force loss, which remains a major drawback [72]. Conversely, the tendon-pulley mechanism can provide higher force by minimising the effects of friction and providing more predictable control [73]. However, the size and weight of this mechanism make it unsuitable for surgical applications. Furthermore, the tendon-pulley mechanism requires the use of pulleys to guide the sliding motion, which adds to the size and mass of the components, especially in devices with complex paths.

People with a cerebrovascular incident or spinal cord injury may experience a reduction in physical strength, paralysis or haemiplegia, which can significantly affect their daily life activities [7, 74]. Functional recovery treatment often involves physical therapy with personal therapists. There is a huge need for robot-assisted rehabilitation that can help maintain therapeutic intensity and task-based exercises in a controllable manner while reducing the workload of therapists, potentially also reducing the cost of treatment [8, 75]. According to clinical studies, robot-assisted devices have been shown to be more effective in improving patient outcomes than conventional physiotherapy [5, 6]. A variety of actuation mechanisms including rigid linkage, tendon-driven, and soft actuation has been proposed to power these robot-assisted devices for upper limb movements and lower limb gait rehabilitation.

2.2.2. Rigid-frame rehabilitation robots

Rigid-frame robotic assist devices for upper and lower limb rehabilitation consist of serially connected rigid links and joints, which can be divided into two subcategories including end-
effector robots and exoskeleton devices [76]. End-effector robots generate motion at their distal end manipulators, which are connected to the patient’s limb to transmit motion. Due to single-point contact and no human-robot joint alignment, these robots cannot effectively control the entire limb [7]. Some of the end-effector robots are Bi-Manu-Track [77], MIME [78] and MIT-Manus (Fig. 2.2A) [79]. Exoskeleton devices have multiple connection points to the patient’s limb with a one-to-one correspondence between the device and human joints [76]. The exoskeleton is mounted along the patient’s limb and can guide the movement of each joint following a predetermined trajectory. Several exoskeleton devices for upper limb assistance are RUPERT (Fig. 2.2B) [80], ARMin III [81] and SUEFUL-7 [82].

Robotic assist devices that specifically target lower limb rehabilitation to help patients in gait training can be categorised into static robots and overground robots [76]. Static robots aid gait training by providing the patients with an exoskeleton robot to assist with leg movement on a treadmill as well as a body weight support system to reduce the body’s gravity on the legs [8]. This type of training is set up in a fixed and confined space. Some of the treadmill-based exoskeleton robots are Lokomat (Fig. 2.2C) [83], LOPES [84], and ALEX [75]. Overground exoskeleton robots equip the patients with a mobile platform (wearable suit) that attaches to the lower extremities to assist the wearer during walking. Through pre-programmed walking patterns, the patients gradually regain a normal gait. The overground exoskeleton often needs the help of the upper limbs to maintain balance while walking. Examples of this category include ReWalk (Fig. 2.2D) [85], MINDWALKER [86] and HAL [87].

2.2.3. Soft wearable robots

Soft wearable robots or soft exoskeletons are becoming increasingly popular due to their lightweight and high conformability. These devices, which replace rigid links and joints with soft and flexible components, can be easily adapted to the patient’s limb, similar to clothing [88]. It is possible to create a portable and wearable soft robotic system due to the conformability and compactness of its components. The tendon-sheath mechanism is attached to fabric substrates to transmit motion and force from a portable actuation unit located at the back or waist to the targeted limbs. For example, Xiloyannis et al. [88] and Chiaradia et al. [89] created a fabric-based soft wearable skeleton, also known as an exosuit, with tendon-driven actuation for supporting elbow movements and hand grasping (Fig. 2.2E). The exosuit helped to reduce the wearer’s muscular effort by 64.5% [89]. Similarly, Quinlivan et al. developed a lower-body exosuit driven by a tendon-sheath mechanism to aid in walking (Fig.
2.2F) [90]. The exosuit applies force to the wearer’s ankles and waist to assist with plantar flexion and hip flexion. Results of the study showed that the metabolic rate decreased by up to 23% for individuals when walking with the exosuit.

**Figure 2.2.** Robot-assisted devices for rehabilitation. (A) MIT-Manus [79]. (B) RUPERT [80]. (C) Lokomat [83]. (D) ReWalk [8]. (E) Exosuit for elbow assistance [89]. (F) Soft exosuit to aid in walking [90]. (G) Exo-Glove [74] and (H) Exo-Glove Poly II [91] for hand assistance.
In another study, In et al. introduced a soft wearable robot for hand assistance called Exo-Glove made by routing tendons around a conventional glove to induce flexion and extension movements of the fingers (Fig. 2.2G) [74]. The Exo-Glove was capable of producing 20 N of pinch force and 40 N of grasping force and assisted in grasping arbitrarily shaped objects. Based on this study, Kang et al. developed the second version of the Exo-Glove called Exo-Glove Poly II that replaced the fabric glove with polymer materials (Fig. 2.2H) [91]. The new design was compact and suitable for multiple hand sizes.

It is clear that soft wearable robots provide many advantages over their rigid counterparts. However, the tendon-driven mechanism used in these devices presents challenges, such as high force loss and nonlinear hysteresis due to friction. Furthermore, the shear force generated by tendons has the potential to fracture the robots and harm the patients [74].

2.3. Wound closure technologies

2.3.1. Clinical need

Wound closure is the process of sealing or closing a wound, which is a break in the skin or tissue, to promote healing, prevent infection and minimise scarring [92]. Wound closure can occur naturally through tissue repair and regeneration or can be facilitated by medical interventions such as sutures and staples for large wounds or those in high mobility areas while adhesive tapes and skin glue may be used for smaller wounds or those in areas where there is less tension. In the case of internal tissue perforation, wound closure with surgical sutures remains the most common practice because of their reliability and versatility.

Flexible endoscopy has gained popularity through diagnosis and considerably expanded its scope to provide therapeutic treatments. Unlike traditional open surgery and laparoscopic surgery, flexible endoscopic surgery brings many benefits to patients. For instance, patients who undergo flexible endoscopic surgery can heal more quickly from the procedure, have less postoperative pain, and do not have any scarring on their abdominal skin [3]. One of the key challenges for flexible endoscopic surgery is the closure of perforations that may occur during surgery [93, 94]. For example, endoscopic submucosal dissection (ESD) has a high risk of perforation, ranging from 5 to 10%, even though this procedure only addresses mucosal tumours [95]. In the case of endoscopic full-thickness resection (EFTR), tissue perforation is inevitable because this procedure involves the resection of the entire tumour, which usually
cuts through the muscular layer [95]. Another instance where an internal perforation is unavoidable is during natural orifice transluminal endoscopic surgery (NOTES). In this procedure, the endoscope accesses the abdominal cavity through an intentional incision in the stomach or colon [93, 96]. This incision needs to be closed after the predetermined procedure is completed. The procedures mentioned above require an effective method to close the perforation.

Effective perforation closure is critical to the safety and success of flexible endoscopic surgery. The edges of the wound should be joined together and secured with appropriate tension. Excessive tension can cause necrosis (tissue death), resulting in pain and other complications for patients. Conversely, if the tension is too little, it can form scar tissue, leading to the possibility of hernia and chronic wounds [97, 98]. Effective wound closure methods will also enable more complex and innovative endoscopic procedures. Advances in perforation closure represent a major shift in the field of minimally invasive endoscopic surgery, which aims to reduce the invasiveness and complications of traditional surgery. As a result, many researchers have explored different methods and technologies to improve perforation closure with a special focus on developing better surgical sutures and suturing devices that can be used endoscopically.

2.3.2. Wound closure technologies

Many studies suggest using endoscopic clips for perforation closure. Endoscopic clips are simple and quick to deploy, which can close wounds through an endoscope without using surgical sutures. They were originally designed for endoscopic haemostasis, which is the process of stopping bleeding from blood vessels. Endoscopic clips are made of biocompatible metallic materials (e.g. medical-grade stainless steel and nitinol) for long-term residence in the human body. They usually have two jaws to clamp tissues together when deployed, creating pressure to stop bleeding [99]. There are currently two types of endoscopic clip concepts based on their size and delivery method. They are through-the-scope clips small enough to fit into endoscope channels, suitable for small-size perforation and over-the-scope clips, which are large clips for haemostasis of severe bleeding and large wounds (Fig. 2.3A) [100].

Another approach for perforation closure is to develop new endoscopic suturing devices that use conventional surgical sutures. Cao et al. developed a robotic suturing system for endoscopic perforation closure (Fig. 2.3B) [101, 102]. This is a flexible endoscopic system constructed with master-slave architecture. The system consisted of a master console to receive
motion commands from the surgeon’s hands and two flexible miniature robotic arms, each with five degrees of freedom for grasping and suturing tasks. With the surgeon’s control, the robotic suturing system was capable of performing running stitches and tying surgical knots endoscopically. The OverStitch endoscopic suturing system (Apollo Endosurgery, USA) is capable of full-thickness suturing with a curved needle for both running and interrupted stitches [103]. The needle passes the suture through the tissue and then deploys two T-tags to secure the closure without tying complex surgical knots. Similarly, the incisionless operating platform (IOP) (USGI Medical, USA) can grab and pierce the tissue to pass a suture through it and then deploy a pair of expandable anchors to close the perforation [104]. In addition, OverStitch can also be used for endoscopic sleeve gastroplasty (ESG) and IOP can be used for primary obesity surgery endoluminal (POSE). These are weight loss treatments that fold stomach tissue to reduce its capacity, thereby promoting weight loss [105, 106].

Apart from developing suturing devices, many studies have investigated methods of upgrading traditional surgical sutures by proposing new designs or using advanced functional materials. Conventional surgical sutures for wound closure often require a secure knot with sufficient strength and compression. However, it is still difficult to tie a knot properly and with enough tension during suturing, especially for endoscopic surgery. Therefore, barbed sutures have been proposed, in which the barbs can anchor themselves to tissues without knotting (Fig. 2.3C) [107, 108]. In terms of material development, a variety of biomaterials with different surface textures, such as silk, linen fibres, polyester and nylon have been employed to create advanced sutures [98, 109]. In recent years, there has been growing interest in functional sutures that can provide some extra benefits to wounds. For example, in addition to joining separated tissues, drug-eluting sutures created by Chen et al. offer drug delivery to the wounds to accelerate the healing process [110]. Likewise, Guyette et al. introduced sutures that were equipped with stem cells to provide biological scaffolds to promote cell growth and tissue regeneration [111]. Liu et al. and Wang et al. developed antimicrobial sutures that were sutures coated with triclosan or silver nanoparticles (these are antimicrobial agents) to prevent infection after the suturing procedure [112, 113].

A new trend for wound closure is the use of smart surgical sutures made of active, programmable and functional filaments. Lendlein et al. introduced a smart suture made from a biodegradable thermoplastic SMP that could tighten a knot when heated [53]. The suture was programmed by elongating to 200% strain and tying a loose knot. The knot was tightened for 20 s when exposed to a temperature of 40°C (Fig. 2.3D). Other smart self-tightening surgical
Figure 2.3. Wound closure technologies. (A) Over-the-scope clip (© OTSC® System, Ovesco Endoscopy AG, Germany, 2014). (B) Master-slave robotic suturing system with two flexible robotic arms [101]. (C) Barbed sutures (© Ethicon US, LLC, 2019). (D) Shape memory sutures provide self-tightening knots when heated [53]. (E) Smart electronic suture to monitor wound temperature [114].
sutures can be found in the reference [115]. SMP-based smart suture technology has the intrinsic drawback of slow response time. It may also be unsuitable for practical use due to the requirement of thermal heating at the surgical site. Several other types of smart sutures act as flexible sensors to monitor wound conditions. Kim et al. embedded flexible microheaters and temperature sensors into polymer strips and used them at the surgical site to monitor wound temperature to aid wound healing (Fig. 2.3E) [114]. There are also other smart electronic sutures with integrated sensors to measure physiological parameters such as pH levels, oxygen levels or wound secretions [114, 116].

2.4. Shape-morphing soft robots and smart textiles

2.4.1. Shape-morphing soft robots

While rigid robots are useful in organised environments, they have difficulty operating in unfamiliar and evolving settings, limiting their use in search and exploration. They are also unsuitable for applications that directly interact with humans, such as wearable assistive devices and surgical instruments. In recent years, a large number of research studies have shifted focus to soft robots for medical applications due to their versatility and conformability. A special branch of soft robotics on smart textiles and shape-morphing soft robots has received considerable attention [117]. These smart textiles and soft robots are flexible, versatile and programmable so they can transform their shapes to adapt to diverse environments, generate force to produce mechanical work and have multiple modes of motion [118, 119]. As such, they have been deployed in many robotic and medical fields, including self-folding robots [120, 121], wearable assistive devices [122], stretchable electronics [123] and biomedical devices [16].

Researchers have been working on developing smart sheets that can be programmed to fold into specific shapes when triggered by external stimuli such as heat or pressure [118]. An effective approach to creating shape-morphing structures involves the combination of various layers of materials with different thermal coefficients of expansion. The difference in the expansion yield of each layer when heated generates bending motion [124, 125]. Lee et al. applied this concept to create a transparent thin film actuator and demonstrated its functions through various soft robots such as a soft gripper, a Venus flytrap and a walking robot [126]. Similarly, Janbaz et al. developed thermally responsive multi-layer strips that could transform into a variety of three-dimensional (3D) structures such as rolled, spiral, self-twisting structures
and self-folding origami (Fig. 2.4A) [127]. Hawkes et al. introduced a programmable self-folding origami sheet made of interconnected triangular sections with embedded actuation [128]. The sheet could fold into predetermined shapes upon activation. Other self-folding origami concepts with integrated stimuli-responsive materials can be found in references [129, 130]. Drawing inspiration from the natural morphogenesis of biological structures, Siéfert et al. introduced baromorphs (pneumatic shape-morphing elastomers) made by strategically arranging airway networks inside an elastic sheet [131]. Upon receiving air pressure, the baromorphs transformed into pre-designed arbitrary 3D structures (Fig. 2.4B).

### 2.4.2. Smart textiles

In another approach, shape-morphing soft robots can be made from traditional textiles or fabrics with integrated actuation mechanisms to induce deformations. Textiles are one of mankind’s oldest technologies, appearing in everyday life from clothing, bed linen and towels to medical applications [29]. Textiles are soft, flexible and conformable sheets made from a variety of fibre materials by interlacing techniques. Textiles have been used to make soft wearable robots, where they serve as passive substrates for attaching other components such as motors, pulleys and tendons and act as intermediate components to transmit motion and force from actuators to human extremities. Examples of this approach include soft wearable gloves to aid hand movements [74, 132] and a lower-body soft exoskeleton to assist with walking [90].

Many studies have implemented textiles as constrained layers to restrict certain movements of soft components to create soft actuators for a wide range of robotic and medical applications. Zhu et al. developed fluidic fabric muscle sheets by using textiles to constrain the radial expansion of silicone tubes to induce multimodal movements such as stretching, squeezing and bending (Fig. 2.4C) [64]. Booth et al. proposed a class of soft robotic skins called OmniSkins made by integrating pneumatic or thermally-driven actuators into fabric substrates (Fig. 2.4D) [133]. The OmniSkins were wrapped around many soft objects to induce desired movements, which have been demonstrated in locomotion robots, soft grippers and wearable devices. Buckner et al. created robotic fabrics by integrating functional fibres such as SMA actuators, variable-stiffness fibres and conductive ink sensors into conventional passive textiles. These robotic fabrics have been deployed in an active wearable tourniquet and deployable structures [134]. Other smart textile concepts can be found in references [135-137].
Figure 2.4. Shape-morphing soft robots and smart textiles. (A) DNA-inspired shape-shifting structure [127]. (B) Pneumatic shape-morphing elastomers [131]. (C) Fluidic fabric muscle sheet produces various motions [64]. (D) Soft robotic skins reconfigure objects from passive to active [133]. E) Smart textiles made of thermally responsive fibres [138].

The latest approach to creating smart textiles is knitting and weaving soft actuators or stimuli-responsive fibres [31, 135, 139]. The motion of participant soft actuators, when triggered by an external stimulus such as electricity, temperature and pressure, generates the motion of their smart textiles. Therefore, this approach provides a high degree of control in terms of motion, shape change and local stiffness. Berzowska et al. developed animated dresses by integrating nitinol wires (SMA actuators) into fabrics [140]. The dress hemline rose and the flowers on the dress closed when heated. Stylios et al. incorporated SMP into woven fabrics for wrinkle recovery [141]. Granberry et al. created a self-fitting wearable sleeve by knitting NiTi-based SMA [142]. The main disadvantages of these thermally-driven smart textiles are slow response time and complicated controls. Maziz et al. knitted and weaved electroactive polymer fibres to
produce electrically responsive smart textiles for wearable applications [30]. Hiramitsu et al. introduced various forms of smart textiles made by interlacing thin McKibben pneumatic actuators and different conventional filaments [143]. The proposed smart textiles could provide moderate strain and high force. Haines et al. developed a new class of thermally responsive artificial muscles made of highly twisted fibres (fishing line and sewing thread) and then used them to produce smart textiles using conventional knitting and weaving machines (Fig. 2.4E) [26, 138].

2.5. Cardiac assistive devices

2.5.1. Clinical need

The heart is a vital organ in the human body. It has the function of circulating blood throughout the body to maintain normal functions of every cell, tissue and organ. The heart is the hardest working organ with a strong muscular structure that beats continuously throughout a person’s lifespan. The heart is literally a biological pump in the circulatory system that delivers oxygen and nutrients to the body as well as removes carbon dioxide and other waste products [144]. HF is a progressive heart condition where the heart is unable to provide enough blood to the body, which can cause heart valve problems, arrhythmias and kidney failure and affect the entire metabolic system [145]. Symptoms may include chest pain, reduced blood supply to vital organs and irregular heartbeats. Potential causes of HF include myocardial infarction (heart attack), high blood pressure, cardiomyopathy and coronary artery disease (CAD). By 2030, HF will afflict around 90 million people globally, including more than 0.75 million Australians with an annual national healthcare cost of $3.8 billion [146, 147].

Treatment of HF depends on the progression of the disease. Mild HF can be treated with medications to relieve symptoms. For example, angiotensin-converting enzyme (ACE) inhibitors help open narrowed blood vessels. Beta-blockers are used to lower blood pressure and slow the heart rate [148]. Severe HF often requires surgical interventions such as coronary artery bypass graft (CABG) to correct blocked coronary arteries and heart valve surgery to repair or replace damaged valves [149]. Heart transplantation is recommended in extreme cases of end-stage HF [150]. However, the availability of donor hearts is limited, leading to the deaths of many patients awaiting transplantation.
Researchers and scientists worldwide are searching for an effective treatment method or procedure to deal with HF, aiming to slow or even cease the disease’s progression from mild to severe. In the field of medical robotics, nanoscale and microscale robots have been developed for targeted therapy and cardiac regeneration for ischaemic heart disease. These miniature robots are employed to transport and deliver specific molecules to the targeted site for the treatment of ischaemic myocardia [151, 152]. Diller et al. developed magnetically-driven micro-grippers for the 3D assembly of micro-parts in remote environments. These microbots have the potential for drug delivery inside the human body [153]. On a larger scale, the development of implantable cardiac assist devices has been proposed to bridge the therapeutic gap between medications and heart transplantation. This type of mechanical support either shares or takes over the workload of the compromised heart to restore the normal function of the circulatory system. The implantable cardiac assist devices can be deployed as an intermediate treatment for bridge-to-transplant patients while waiting for an available donor heart or can also be served as destination therapy for those who are not eligible for heart transplantation. The following section introduces three different types of mechanical implantable devices for cardiovascular support.

2.5.2. Ventricular assist devices

A ventricular assist device (VAD) or mechanical circulatory support device is a mechanical pump that circulates blood from ventricles to the entire body through pulmonary, systemic and coronary circulations [154]. There are three different types of VAD based on the implanted ventricle, including (1) left VAD (LVAD, implanted in the left ventricle (LV), the most common type) to pump oxygenated blood from the LV to the aorta and branch out the whole body, (2) right VAD (RVAD) to pump deoxygenated blood from the right ventricle (RV) to the pulmonary artery and the lungs and (3) biventricular assist device (BiVAD) to support both ventricles at the same time. A typical VAD consists of a mechanical pump to generate motion to move blood, an inflow cannula connected directly to the ventricle to receive blood flow and an outflow graft that connects the mechanical pump to the corresponding blood vessels (aorta, pulmonary artery, or both). In addition, the VAD also has a control unit comprising a control board and batteries located outside the body and a transcutaneous driveline cable to connect the mechanical pump to the control unit through the patient’s skin. The configuration of the VADs allows them to take over the pumping function of the heart. VADs have been proven to enhance cardiac function and improve the quality of life in patients with advanced HF [155].
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There are many commercially available VADs in the market. The HeartWare HVAD System (Medtronic, USA) features a centrifugal flow pump that connects directly to the bottom of the LV [156]. The HeartWare provides a continuous flow capable of supplying sufficient blood to relieve symptoms of HF [157-159]. However, the device has been ceased for use since Jun 2021 because of an increased risk of fatal and neurological adverse events associated with the internal pump and the possibility of delay or failure to restart the internal pump [160]. The LifeSPARC system (LivaNova, USA) offers ready-to-deploy kits to provide temporary support to patients during high-risk cardiac procedures. The system features an extracorporeal ventricular assist device called TandemHeart to provide partial or total cardiopulmonary bypass during open surgical procedures on the heart or great vessels or temporary circulatory bypass for open surgical procedures on the aorta or vena cava [161, 162]. The TandemHeart consists of a drainage catheter inserted into the left atrium through the femoral vein and perforation through the septum, a centrifugal pump for blood circulation and a return cannula connected to the femoral artery to return blood to the systemic circulation [162]. The HeartMate 3 (Abbott, USA) is an LVAD for short- and long-term mechanical circulatory support in adult and paediatric patients with advanced refractory LV HF [163, 164]. This is an implantable device with a similar structure to the HeartWare. However, the HeartMate 3 features the full magnetic levitation technology that eliminates hydrodynamic or mechanical bearings, reduces shear stress and provides intrinsic pulsatility that minimises thrombosis [165, 166]. Other VADs include Jarvik 2000 [167, 168] and Heart Assist 5 [169, 170]. A comprehensive review of mechanical circulatory support systems can be found in the references [154, 171].

Taking a different approach, the BiVACOR total artificial heart (TAH) (BiVACOR, USA) is designed to fully replace the function of the natural heart for long-term implantation in patients with end-stage HF [174-176]. The BiVACOR TAH has two centrifugal impellers mounted on a single rotor to simultaneously provide balanced pulsatile outflow to the systemic and pulmonary circulations. The device features magnetic levitation technology to provide precise, stable operation with no mechanical wear. In addition, the smart controller can adjust pump operation to changes in patient activity with outflow capacities of up to 12 L/m.

Although VADs help reduce symptoms of HF and improve quality of life, their use is associated with many potential complications such as bleeding, infection and thromboembolism from direct blood contact with nonbiological device components [154, 172, 173]. In addition, long-term use of blood-thinning medications is required to prevent the formation of blood clots.
2.5.3. Passive ventricular constraint devices

Passive ventricular constraint devices (PVCDs) for the treatment of HF are based on the principle of limiting heart enlargement using a thin layer of flexible and passive materials [174]. PVCDs are designed to be implanted on the heart’s exterior epicardial surface to provide ventricular support that reduces stress on cardiac walls in patients with dilated cardiomyopathy. In addition, PVCDs help to reduce myocardial stretch and end-systolic volume and interrupt periodic eccentric hypertrophy, leading to the cessation of LV remodeling [175, 176]. Several cardiomyoplasty studies have shown that LV performance can be improved by preventing further dilation using passive diastolic constraint [177, 178].

This passive constraint concept has been applied to create cardiac support devices (CSDs) for HF treatment. For example, the Acorn CorCap CSD (Acorn Cardiovascular, USA) is a knitted mesh made of polyester fibres to enclose the heart covering both the left and right ventricles to limit its dilation [175]. The results of a five-year follow-up randomised trial demonstrated the long-term safety of CorCap CSD implantation and sustained LV reverse remodeling [179]. The Paracor HeartNet (Paracor Medical, USA) is a ventricular constraint device made of silicone-coated nitinol wires arranged in a meandering path to facilitate radial expansion [175]. The HeartNet is implanted around the heart to provide epicardial pressure, thereby reducing stress on the ventricular wall and potentially reversing remodeling. Clinical studies showed a high rate of transplant success (98%) in patients with HF as well as initial benefits toward reverse remodeling [180, 181]. Taking a slightly different approach, Ghanta et al. introduced an adjustable ventricular restraint device that has a half-ellipsoidal balloon shape made of medical-grade polyurethane sheets [182]. The contact pressure of the device on the heart can be adjusted to obtain an optimal treatment result. The study concluded the dependence of several physiological parameters including the rate of LV reverse remodeling on the degree of ventricular restraint and suggested periodic adjustment of restrain level to optimise therapy efficacy [182].

Treatment of HF with PVCDs has the advantage of eliminating adverse events caused by direct blood contact with the devices. PVCDs provide a passive restraint to the heart to prevent further ventricular dilation rather than active support for compression. Therefore, most PVCD treatments yield modest improvements in HF symptoms [183, 184]. These devices are preferred for use in the treatment of early-stage HF to prevent disease progression.
2.5.4. Direct cardiac compression devices

Direct cardiac compression devices (DCCDs) have been developed to address the limitations of VADs (e.g. blood contact) and PVCDs (e.g. passive support). Amongst mechanical devices for HF treatment, DCCD is the most advantageous method in terms of high therapeutic efficacy, fewer complications, and therefore a safer option than conventional VADs [150, 172, 173, 185, 186]. DCCD treatment is classified as active therapy that can provide direct compression to the exterior (epicardium) of the failing heart to treat end-stage HF [187]. The DCCDs share the pumping workload with myocardial muscles so that they can work together to restore the heart’s normal ability to deliver adequate blood flow to the body. The DCCDs are usually cup-shaped to facilitate mounting to the cardiac surface and are reinforced with sutures or vacuum pressure. Similar to PVCDs, DCCDs cover the epicardial surface of the heart, thereby eliminating adverse events related to blood contact. The operation of DCCDs involves sequentially switching between two phases, including the compression phase to compress the heart to pump blood out of the heart ventricles and the relaxation phase to allow passive filling of blood into the atria and ventricles. An electrocardiographic (ECG) controller is required to synchronise the device’s movements with the heart.

The first DCCD was introduced by Bencini and Parola in 1956 called pneumo-massage [188]. The authors used air to inflate and deflate the pericardial space to compress and decompress the heart. This method was ineffective but inspired further development. Anstadt et al. developed the Anstadt cup for patients with cardiac arrest [189]. The device is designed to surround the circumference of the heart and consisted of a firm outer layer and an inflatable inner layer that is actively controlled to assist both systole and diastole. Clinical studies have demonstrated the restoration of cardiac function with sufficient blood pressure and cardiac output and significant improvement in haemodynamics as well as quick implantation procedure [190, 191]. The AbioBooster (AbioMed, USA) is a pneumatic DCCD for long-term implantation to treat HF [192, 193]. The device consisted of a series of inflatable polyurethane tubes coated with silicone and designed to enclose both ventricles of the heart. The AbioBooster could actively aid both systole and diastole by pressurising and depressurising its inflatable tubes respectively. It was capable of producing an aortic pressure of 115 mmHg and a blood flow rate of 6.5 L/min. The HeartPatch (Heart Assist Technologies, Australia) is a pneumatic DCCD consisting of two independent inflatable silicone patches bio-integrated into the epicardium of both ventricles using a porous silicone material as the adhesive medium [194]. The patches are actively controlled to inflate and deflate by an extracorporeal pneumatic...
pump and synchronised with heart movements by ECG. Two air chambers of the patches can be independently actuated with adjustable actuation pressure to facilitate different treatment strategies such as univentricular (LV or RV) and biventricular assistance [172, 195]. The HeartPatch approach has potential drawbacks including the risks of myocardial rupture and delamination from the epicardial surface [187].

Figure 2.5. Cardiac assist devices. (A) HeartWare ventricular assist device (Medtronic, USA) [154]. (B) Acorn CorCap cardiac support device (Acorn Cardiovascular, USA) [174]. (C) HeartPatch direct cardiac compression device (Heart Assist Technologies, Australia) [172]. (D) Soft robotic sleeve mimics cardiac muscles [196]. (E) CorInnova epicardial compressive heart assist device (CorInnova, USA) [197].
Advances in soft robotic technology have expanded the arsenal of actuation mechanisms to construct novel DCCDs for HF treatment. SAMs with their flexible, versatile and conformable characteristics are beneficial in providing proper integration with the epicardial surface while inducing desired movements to actively assist the failing heart. Roche et al. introduced a soft robotic sleeve that can be implanted around the heart and provides simultaneous compression and twisting motions to the heart [196, 198]. The sleeve consisted of two layers of silicone elastomer (Ecoflex 00-30) with embedded PAMs arranged in circumferential and helical directions. Both layers are independently controlled to generate compression and twist, mimicking the natural motion of the human heart. However, the sleeve is bulky and difficult to deploy, requiring a large incision for implantation. It also requires a complex, bulky, and noisy source of compressed air to operate. Kongahage et al. developed silicone-coated electrothermal actuators and implemented them into a new VAD that was capable of providing pulsatility control [199]. The study also introduced the concept of an artificial heart sleeve that surrounds the heart to support its function. The CorInnova epicardial compressive heart assist device (CorInnova, USA) is intended for the treatment of acute HF [200]. The device has two semi-ellipsoidal layers of silicone separated by helically arranged nitinol wires. The inner layer is filled with fluid to create highly conformable contact with the epicardium while the outer layer is pneumatically actuated to actively assist the heart during systole. The CorInnova device is deployable so it can be implanted with minimally invasive intervention. It provides different modes of assistance including active cardiac compression to enhance cardiac output and passive constraint to prevent ventricular dilation [197].

2.6. Summary

This chapter has introduced the current development of SAM technologies and four potential fields of interest in robotics and medicine for the implementation of the proposed artificial muscle. The advantages and limitations of these technologies and devices were also highlighted.

Overall, SAMs have advantages over rigid actuators in terms of flexibility, conformability, resilience to disturbances and human-friendly interactions. These characteristics make them suitable for many emerging applications such as wearable assistive devices, smart garments, cardiac assist devices, soft surgical instruments, soft grippers, biomimetic soft robots and soft locomotion systems. However, SAMs often have limited force generation and durability. Their
movements are difficult to sense and monitor, leading to the complexity of precise and reliable control.

There are broadly four main types of SAMs based on their source of excitation. Electrically-driven SAMs are typically responsive and have high strain, but have limited force exertion, low durability (in the case of IPMCs) and require high operating voltage (in the case of DEAs). Magnetically-driven SAMs offer programmable movements and fast response time but generate weak force and require bulky external driving sources. Thermally-driven SAMs can provide high strain and stress but have the drawbacks of slow response time and low energy efficiency. Pressure-driven SAMs are responsive and can generate high force but have limited strain (in the case of McKibben PAMs) and require bulky pneumatic compressors to operate. There is a need for a flexible, responsive, multifunctional SAM with high energy efficiency for a wide range of robotic and medical applications.

There is a paradigm shift from rigid-frame robots to soft wearable devices for rehabilitation in patients with a cerebrovascular incident or spinal cord injury. Although rigid-frame robots are powerful and easy to control, they are bulky and heavy, and their interaction with human limbs remains a major safety concern. In contrast, soft wearable robots are lightweight and highly conformable, which can provide human-friendly interaction. However, current soft wearable devices rely on the tendon-driven mechanism as a means of force transmission, which has the limitations of high force loss, nonlinear backlash hysteresis and potential harm to the patients from the shear force generated by tendons.

Effective wound closure remains a critical challenge to the safety and success of surgical intervention, especially flexible endoscopic surgery. Endoscopic clips offer rapid deployment but can potentially damage tissues due to their strong clamping force. Many advanced flexible endoscopic systems have been developed to perform perforation closure using traditional surgical sutures. A great effort has been made towards the development of functional and smart surgical sutures to prevent infection, monitor wound conditions and accelerate wound healing. A limited number of studies have investigated the use of soft actuators to create automatically tightened smart sutures. Existing thermoplastic SMP sutures could tighten tissues when heated but have a slow response time.

Research interest in shape-morphing soft robots and smart textiles for wearable assistive devices, stretchable electronics and biomedical devices has grown rapidly. A variety of design concepts have been proposed, ranging from thermally-responsive multi-layer strips to
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Pneumatic shape-morphing elastomers to textile-based soft wearable devices and soft textile actuators. In a recent approach, smart textiles are created by knitting and weaving long SAMs to provide multimodal deformation. This method offers great control over local stiffness and shape change. However, challenges remain regarding the performance of the SAMs themselves as well as their ability to form united planar structures. A thin, long, flexible and versatile SAM is required to extend the reach of smart textiles for practical applications.

Heart failure is a global epidemic. Mechanical heart assistive devices provide life-saving treatment for patients with advanced HF, bridging the gap between medication for mild HF and heart transplantation for end-stage HF. Ventricular assist devices and total artificial hearts are mechanical pumps that take over the pumping function of the native heart to circulate blood. They can provide sufficient blood flow to the body, thereby reducing HF symptoms. However, these devices come into direct contact with circulating blood, potentially causing infection, bleeding and thromboembolism, as well as the need for long-term use of blood thinners. Passive ventricular constraint devices provide passive restraint to prevent further ventricular dilation. Their use eliminates the side effects of blood contact but yields modest improvements in HF symptoms. Direct cardiac compression devices feature soft, controllable structures surrounding the cardiac surface to actively provide compression to assist the heart in pumping blood out of the ventricles. Despite advances, some devices are large with thick walls, making implantation difficult while others require bulky pneumatic pumps and complex control systems to operate. Most direct cardiac compression devices cannot mimic the natural motion of the heart, limiting their effectiveness in the treatment of HF.

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Hydraulic filament artificial muscles

This chapter presents the fundamental development of a soft hydraulic filament artificial muscle – a type of pressure-driven soft actuator made of silicone tubing and an extension coil spring. In detail, the design, fabrication, operating principles, and mathematical models of the new artificial muscle are investigated. The proposed artificial muscle has a high length-to-diameter ratio and is fabricated using a simple and low-cost fabrication method, allowing for scalability and mass production. Various aspects of the proposed artificial muscle characteristics including elongation and force exertion corresponding with input volume and pressure, frequency response, durability, elongation limit, energy efficiency, and force tunability, are experimentally examined. Furthermore, this chapter presents the potential implementation of the proposed artificial muscle to replace the more traditional tendon-pulley and tendon-sheath mechanisms. The proposed artificial muscle is demonstrated to actuate a wearable soft robotic glove to assist in grasping and holding various objects.

This work has been published:


The foregoing publication is used in lieu of this chapter, meaning that no modifications have been made to the text. However, the formatting and numbering of subtitles, figures and tables are changed for readability. This disclaimer also applies to subsequent chapters.
3.1. Abstract

The use of soft artificial muscles (SAMs) is rapidly increasing in various domains such as haptics, robotics, and medicine. There is a huge need for a SAM that is highly compliant and facile to fabricate with performance characteristics similar to human muscles. This paper introduces bio-inspired soft hydraulic filament artificial muscles (HFAMs) that can be extended and contracted under fluid pressures. The HFAMs, which have a high aspect ratio of at least 5000, use a simple and low-cost fabrication method of insertion, enabling scalability and mass-production while increasing its generated force via a stiff constrained helical layer and an adjustable stretch ratio of their inner silicone microtube. The developed muscles can produce a high elongation of 246.8% and a high energy efficiency of 62.7%. In addition, the HFAMs can generate a higher contraction force compared to existing state-of-the-art devices via their constrained hollow layer and the adjustable stretch ratio of the inner microtube, enabling a tunable force capability. Experiments are carried out to validate the HFAM performance including durability, lifting, frequency response, and energy efficiency tests. The HFAM capabilities are demonstrated via various experiments, offering a potential substitute for the conventional tendon-driven mechanisms with less friction loss and stable energy efficiency while working against long and tortuous paths. A HFAMs-driven soft exoskeleton glove that could assist in grasping multiple objects is developed and evaluated. The new muscles open great opportunities for research and commercial sectors including emerging applications such as soft wearable devices and flexible surgical robots.

Index terms: fluid-driven artificial muscles, soft actuators, soft exoskeletons, soft robotics, tendon-driven mechanisms, wearable devices.

3.2. Introduction

Robot-assisted surgeries have emerged in the surgical field since 1985. They have become increasingly prevalent for healthcare delivery worldwide [1, 2]. Surgical robots offer great benefits to patients including safer procedures, shorter recovery time, and lower risk of complications compared to conventional surgeries [3]. Tendon-driven mechanisms are dominant in most surgical robots due to their small size and lightweight [4]. In the tendon-driven mechanisms, the desired position and force are transferred from external actuators to distant joints via either rotational pulleys (tendon-pulley mechanism) or routing flexible tubes (tendon-
sheath mechanism). The tendon-pulley mechanism can provide high force transmission because of low friction loss. However, its construction is well-suited for rigid laparoscopy instruments. In contrast, the tendon-sheath mechanism demonstrates a better ability to adapt to unpredictable bends inside the human body such as the gastrointestinal (GI) tract, and has been broadly employed in many surgical robotic systems and flexible surgical devices [5]. However, its inherent drawbacks include high force loss and low speed due to nonlinear friction and backlash hysteresis between the sliding tendon and the routing sheath [6].

Depending on the construction of surgical instruments and how they are being introduced into the human body, there are two main streams of robot-assisted surgical techniques: (i) Robot-assisted laparoscopy for minimally invasive surgery (MIS) with well-known systems such as Da Vinci® surgical systems [7], Senhance® surgical system [8], and Versius® surgical robotic system [9]; (ii) Flexible endoscopy for natural orifice transluminal endoscopic surgery (NOTES) with the MASTER system [5], Flex® robotic system [10], and NeoGuide colonoscope [11].

Patients who suffer from a stroke or spinal cord injury typically experience a loss of physical strength, paralysis, or hemiplegia, directly affecting their daily life activities [12, 13]. To receive medical treatment, these patients are normally prescribed physiotherapy for rehabilitation and functional recovery. Robot-assisted rehabilitation can reduce the workload of therapists while maintaining high therapy intensity, repetition, and task-oriented exercises in a controllable manner [14, 15]. Clinical studies have demonstrated that robotic assistive devices can provide patients with better outcomes than those treated with conventional physical therapy [16, 17]. These assisted devices are typically powered by electrical motors to transfer motion and force to participating links and joints (either rigid or soft). This powerful yet compact tendon-driven mechanism offers high adaptability to various exoskeleton systems ranging from upper limb movements to lower limb gait rehabilitation.

Recently, advanced actuation technologies have emerged to develop soft exoskeletons that are lightweight, compact, and highly conformable. These devices have been demonstrated as useful candidates to replace rigid links and joints. They comprise soft, fabric-based elements that can be easily adapted to the patient’s limb, in a manner similar to wearing human clothing [18]. In these systems, motion and force transmission to the targeted limb are generated via a compact tendon routing system which is driven by electrical motors located in a patient-worn back- or waist-pack. Quinlivan et al. [19] introduced a soft wearable exoskeleton (exosuit) for walking assistance where the moment to the wearers’ ankle is directly controlled by a flexible tendon
sheath mechanism. This system demonstrated a reduction in metabolic rate while walking. Lorenzo’s group presented a soft exosuit for elbow joint assistance driven by two Bowden cables [18, 20, 21]. The exosuit which was made from highly conformable fabric and tendon-driven mechanisms was wrapped around the human arm with the tendon actuation source located in the wearer’s backpack, creating a completely portable device [20]. Inspired by the human musculoskeletal system, Exo-Glove [13] and Exo-Glove Poly II [22] developed by the Kyu-Jin Cho group are soft wearable robots for hand and fingers assistance. In these systems, flexible tendons were routed around the index and middle finger at the palm and back sides to actively exert flexion and extension motions. The Exo-Glove can provide a pinch force of 20 N, a grasping force of 40 N, and can grasp various objects with different shapes and sizes [13]. Although soft exoskeletons offer many advanced features compared to their rigid frame counterparts, challenges remain with the tendon-driven mechanism, including high force loss and nonlinear hysteresis due to friction and shear force that may break the device and cause harm to the patients [13]. In addition, the hysteresis profile of the tendon-sheath mechanism is varied with its routing path, resulting in difficulty with precise control of both motion and force [23-25].

Soft robotics research has been actively expanding over the last two decades, attracting great attention from researchers, scientists, research groups, and dedicated journals worldwide [26, 27]. Soft artificial muscles (SAMs) or soft actuators belong to a new branch of actuation modes, in which they are made from soft and compliant materials [28, 29]. In contrast to their rigid counterparts, soft actuators possess several advanced features such as flexibility, versatility, resilience to disturbances, adaptability to dynamic environments, and human-friendly interaction compared to others [29-31]. SAMs can exert motions and forces by various excitation sources such as electrical and magnetic fields [32-36], thermal power [37-39], and fluid pressure [31, 40]. Electrically-driven soft actuators have been a focus because of ease and versatile modulation from electric signals, as well as the fast growth of electrically responsive materials. Electroactive polymers (EAPs) are lightweight, highly flexible, and compliant materials that exhibit moderate strains to achieve relatively high energy efficiency [41]. Dielectric elastomer actuators (DEAs) are non-ionic EAPs driven by electrical fields that can produce at least 100% strain and high-speed actuation [42, 43]. However, this actuator requires a high operating voltage (several kilovolts), potentially causing electrical breakdown and introducing safety concerns for human use. Although ionic polymer-metal composites (IPMCs) require low driving voltage (few volts), they typically possess slow response times, high cost, and low durability, especially when operating in dry air [41]. Electrically-driven artificial muscles based on gels or phase-change
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materials can generate large volume changes, high strain (900%) [44], and high force generation (up to 4.5 kN) with low operating voltage [45]. However, their intrinsic drawbacks include slow response times and low energy conversion efficiency [44, 46].

By introducing magnetic particles inside elastomer substrates, magnetically-driven soft actuators can provide programmable, multimodal, and high-speed motion for soft robotic structures under controlled external magnetic fields [35, 47]. Despite advances, their inherent limitations include weak actuation force and bulky external driving source. In addition, they are greatly affected by the surrounding environment, especially in magnetic environments or when ferromagnetic materials nearby [48]. Shape memory polymers (SMPs) and shape memory alloys (SMAs) are thermally driven actuators that have great performances in terms of high elongation (up to 1000%) [49] and high stress (up to 200 MPa) [50] but suffer from slow actuation speed, high hysteresis, and low energy efficiency. Recently, SAMs made from twisting inexpensive fishing lines and sewing threads have been shown to produce a 49% contraction stroke at 1 MPa load and exert up to 140 MPa stress at 4.5% contraction [51]. Although these high strength polymer fibers possess remarkable performances, they have intrinsic disadvantages similar to those of the temperature-driven actuators with slow response time and low energy efficiency, limiting their use in many applications [52].

Pressure-driven soft actuators, either by pneumatic or hydraulic sources, are the most popular used artificial muscles due to their low cost, simplicity, and superior performance, as well as high strain, high stress, and high energy efficiency [30, 53]. The most well-known SAM in this category is the McKibben actuator, which consists of an outer sleeve with a braided mesh structure and an inner elastic bladder [54, 55]. When a positive pressure source is supplied, the McKibben muscle is radially expanded to exert a longitudinal contraction force along its axial direction. Kurumaya et al. [55] demonstrated that a thin McKibben muscle could produce 15.77 N of contraction force and 34% elongation under 0.5 MPa air pressure. Multiple McKibben muscles can be braided to increase its contraction limit [56, 57].

Inspired by the McKibben muscle, Belding et al. [58] recently developed a new concept of fluid-driven muscles, in which the outer mesh was replaced by a plastic tube that was pre-programmed in slits. This type of actuator provided controllable, multiple motions, and sequential actuation. In other attempts, Hawkes et al. [59] proposed an inverse pneumatic artificial muscle (IPAM) that consisted of an inner elastic tube and an outer layer of reinforcing fiber. The IPAM could produce over 300% strain when pressurizing and exert 419 kPa stress when depressurizing.
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Instead of using positive pressure, vacuum power has been explored to generate contraction motion and force for artificial muscles. Yang et al. [60] introduced vacuum-actuated muscle-inspired pneumatic structures (VAMPs) that produced linear motions by applying negative pressure to elastomeric beams. VAMPs can store and release elastic energy with elongation up to 45% and an actuation stress up to 65 kPa. Li et al. [40] proposed a fluid-driven origami-inspired artificial muscle (FOAMs) that produced at least 90% strain, 600 kPa stress, and 2kW/kg power density by vacuum pressure. Zhu et al. [61] developed a new fluidic fabric muscle sheet (FFMS) that integrated fluidic transmissions based on a single tube or arrays of elastic tubes embedded into flexible fabric layers via the use of facile apparel engineering methods of computerized sewing techniques. The FFMS can strain, squeeze, bend, and conform to hard or soft objects of arbitrary shapes or sizes, including the human body.

Although significant progress for soft actuators has been achieved, there is a huge need for artificial muscles that are scalable, have high aspect ratio with simple and mass-produced fabrication methods, and can serve as an active filament for different applications. In this paper, we introduce a hydraulic filament artificial muscle (HFAM) that is scalable, flexible and has a high-speed actuation, high energy efficiency, and high aspect ratio (Fig. 3.1). The new HFAM is created from a simple insertion fabrication method, which avoids complex manual wrapping of an inextensible fiber alongside the muscle. This new method provides a uniform distribution of a helical structure around an inner soft silicone microtube of the HFAM, preventing unexpected failure enabling large elongations at high speeds during actuation. Therefore, it eliminates ballooning during the operation as well as overcoming the limitation on the muscle length. In addition, it provides a new force tunability by varying the stretch ratio, which is not available in existing fiber-reinforced SAMs. The new HFAM enables new means of efficient actuation, similar to those of tendon-driven mechanisms, for use in surgical tools or soft wearable devices.

The subsequent contents are structured as follows. Section 3.3 presents the development of the HFAMs including their design, fabrication, modeling, and characterization. Section 3.4 investigates the HFAM performance as tendon-driven mechanisms, in which both HFAM-like tendon-pulley and tendon-sheath configurations are demonstrated. In section 3.5, we produce an HFAM-based soft exoskeleton glove to demonstrate the use of our muscle in a specific embodiment. The new glove, which can be worn by a user, provides assistance to the wearer for grasping multiple objects of different sizes and shapes. Finally, the discussion, conclusion, and future work are presented in section 3.6.
3.3. Development of soft hydraulic filament artificial muscles

3.3.1. Fabrication method and material choice for a high aspect ratio HFAM

The HFAM consists of two primary components, an inner soft microtubule, and an outer helical coil. Both components are scalable and versatile in terms of sizes and materials, but they must be relatively paired with each other to form an assembly. The soft microtubule can be made from a wide range of materials such as silicone rubbers, Ecoflex™ series (Smooth-On, Inc., Macungie, PA, USA) and silicone elastomer, NuSil™ (NuSil™ Technology LLC, Carpinteria, CA, USA) using molding or rolling coating processes. In the molding technique, uncured silicone is prepared and then poured into a long cylindrical mold with a center micro rod while rolling coating requires uncured silicone to be coated outside a continuously rotating center rod [62] (Fig. 3.2A). The curing time of the silicone can be shortened by heating. The soft microtubule is subsequently obtained by discarding the mold and the center rod. Parameters of the tube, such as outer diameter, inner diameter, and length, are defined by the specifications of the mold and the center rod. The obtained silicone tube has the properties of its substance. For instance, Ecoflex™ 00-30 has an elongation at break of 900%, 1.38 MPa tensile strength, and shore hardness of 30.
Another approach to achieve a soft microtubule is off-the-shelf tubing. A huge collection of silicone tubing is available at Saint-Gobain S.A. (Courbevoie, France) and McMaster-Carr Supply Co. (Elmhurst, IL, USA). They are industrial-made products with numerous specifications and material properties.

**Figure 3.2.** Illustrated fabrication processes of the soft microtubule and helical coil. (A) Soft microtubule; (1) Uncured silicone elastomer (Ecoflex™ or NuSil™) is coated on a base plate by spin coater or film applicator; a rod which is secured in the chuck of a power drill is coated with a layer of uncured silicone; (2) Heating the rod while rotating to speed up the curing progress; (3) Removing the rod to obtain the microtubule. (B) Helical coil; (1) A winding machine is used to provide mandrel rotation while a wire guide is fed by a wire (stainless steel, brass, or fishing line) to form the helical coil; (2) Heating the coil to relieve its stress and stabilize its shape; (3) Removing the mandrel and grinding both its ends to obtain the helical coil. (C) Making a fishing line coil (polyvinylidene fluoride: PVDF) with the help of a carbon fiber rod as a mandrel, power drill, and heat gun (left panel) and the obtained fishing line coil (right panel).
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One of the main advantages of our HFAM is the use of a separate constrained outer layer, a type of long helical coil that can be manufactured from a diversity of inextensible materials such as stainless steel wires, brass wires, fishing line, or sewing thread. A winding machine that provides a rotational motion of a mandrel and longitudinal translation of a wire guide simultaneously is required to produce a helical coil. Figure 3.2B shows the fabrication process for the HFAM helical coil. After winding the wire around the mandrel to reach the desired length, the acquired coil should be placed in an oven (in case of fishing line) or using a heater (in case of stainless steel or brass wire) to relieve stress and stabilize the coil shape. The stiffness and dimensions of the obtained helical coil are affected by its material properties, wire diameter, and the mandrel’s size. Theoretically, the stiffness of the helical coil corresponds with the wire diameter and is inversely proportional to the mandrel diameter.

As an example, Fig. 3.2C shows the fabrication process for the helical coil made from fishing line. First, the fishing line (polyvinylidene fluoride: PVDF, 0.2 mm diameter) is wrapped around a carbon fiber rod with an outer diameter of 1.5 mm (CST - The Composites Store, Inc., CA, USA) via a power drill and vice. Then, a heat gun generated hot air at around 80°C pointing and longitudinally moving along the rod while it was rotating by a power drill. These two simultaneous motions enable even distribution of the heat to the fishing line and therefore stabilizing the coil shape. The power drill was running at 60 rpm (round per minute), and the heat gun linearly moved at around 5 mm/s back and forth twice. The last step was removing the carbon fiber rod to obtain the helical coil with an outer diameter of 1.95 mm and a length of 400 mm. Alternatively, the carbon fiber rod with helical fishing line wrapping outside could be placed inside an oven at 80°C for 6 minutes to achieve a similar result. Similar to the soft microtubule, the helical coil is commercially available for purchase.

To obtain the HFAM, the soft microtubule is directly inserted into the hollow constrained coil (Fig. 3.3A) using a sewing thread or fishing line serving as a guide that is attached to a needle or a long carbon fiber rod. Once the soft microtubule is completely inserted into the coil, one end is tied into a knot and permanently adhered to the coil end by adhesive glue (LOCTITE®, USA). In contrast, the other end is connected to a commercial fluid transmission tube (Cole-Parmer, USA), which is connected to a fluid control source. The developed HFAM can receive power sources from a variety of transmission means, including compressible air or gas or incompressible fluids such as water, saline, or oil. This fabrication method can be used to produce HFAMs with any predetermined lengths and sizes. In this paper, we fabricated a long HFAM and then stored it in a spool that can be cut to meet a specific length requirement afterward, similar to how we use the
Figure 3.3. HFAM assembly and prototypes. (A) Insertion method enables HFAM to be produced at a meter-long scale with a high aspect ratio. (B) HFAM prototypes; (1) Silicone rubber microtubule and fishing line coil (PVDF); (2) Silicone rubber microtubule and stainless steel coil; (3) Latex rubber microtubule and stainless steel coil; (4) The muscle in (3) after coating a layer of Ecoflex™; (5) Latex rubber microtubule and stainless steel coil. (C) HFAM with a high aspect ratio of 5000 made from soft silicone rubber microtubule and stainless steel coil.

spool of electrical wires or silicone tubes in daily life. This contrasts with the conventional approach with fiber-reinforced soft extensible actuators where an inextensible fiber is manually wrapped around a soft elastic tube. The primary fabrication presented in this work leverages the simplicity of an insertion method, in which an inner elastic microtubule is directly routed through an outer constrained helical coil. This method allows facile manufacturing of miniature, meter-
long muscle with a diameter ranging from a few hundred micrometers to several millimeters. It also avoids the strict requirement of uniform fiber wrapping along the soft elastic microtubule, which poses many challenges for any fluid-driven soft actuators at the micrometer scale with high aspect ratios.

We demonstrated the scalability and versatility of our HFAM by producing diverse prototypes in various sizes and material combinations (Fig. 3.3B). The soft microtubule can be either silicone elastomer or latex rubber. The helical coils are stainless steel or fishing line. We also demonstrated the high aspect ratio property by fabricating a miniature, meter-long HFAM from a silicone elastomer microtubule and stainless steel coil with an outer diameter of 0.8 mm and a length of 4 m (OD 0.8 mm x L 4000 mm, Fig. 3.3C and Supporting Video). The obtained HFAM has a small size and large aspect ratio of 5000:1 that is extremely high compared to existing hydraulic or pneumatic SAMs reported in the literature [40, 59, 63].

3.3.2. Operating principle of the HFAM and modeling

One of the main advantages of HFAM is the capability to transmit mechanical force or energy from a distance, similar to that of the conventional flexible tendon or cable mechanisms in which one end is fixed and the other end moves. HFAM-based tendon-driven mechanisms can exist as two main types: (i) an HFAM-based tendon-pulley mechanism where the HFAM directly conveys the force and motion without using any hollow sheaths; and (ii) an HFAM-based tendon-sheath mechanism where an inner HFAM slides inside a hollow sheath (Fig. 3.4A). Compared to conventional tendon-driven systems, both types of HFAM offer higher energy efficiency due to the use of a fluid source, allowing its external power supply to be located at a long distance without sacrificing its input energy. In addition, the HFAM generates its motion via a local extension of an individual muscle segment, which is analogous to the natural behavior of certain plant cells that are lengthened or shortened when being pressurized [64], enabling uniform distributions of the motion and its generated mechanical force while maintaining its energy regardless of the distance. In contrast, the motion and force output of the tendon-driven mechanism highly depends on the routing path and the distance of its power source located at its proximal end [65, 66]. In applications that require highly tortuous paths and multi-loop configurations such as flexible surgical robots or wearable exoskeletons where tendon-sheath or Bowden cable mechanisms are currently dominant, a high force loss is unavoidable [2, 67]. Our HFAM-based tendon routing sheath mechanism, in contrast, overcomes this limitation with a minimal energy loss.
Figure 3.4. Two operating modes and analytical illustration. (A) The HFAM can function as a tendon-driven mechanism with or without an outer sheath. (B) Generated motion and force diagram of an HFAM in relation to the applied input pressure.

We describe here the detailed working principle of the HFAM, mainly focusing on the HFAM mechanism without the sheath. The working principle of the HFAM with a sheath is similar, except it is routed through the inner channel of a guiding sheath, allowing a multi-loop, highly flexible, and complex path to convey the force and motion. In the HFAM, the outer helical coil restrains the radial expansion of the soft microtubule, leading to a lengthwise enlargement of the muscle. The HFAM generates a contraction force $F_{\text{out}}$ by switching between low-pressure and high-pressure phases. Specifically, it stores elastic potential energy (EPE) by elongating to a certain length, then it converts this EPE to mechanical power by reducing the applied pressure. For the HFAM without the sheath, depending on the nature of specific applications, each HFAM can be arranged in a certain configuration, material combination, and predetermined parameters such as outer diameter and original length $l_0$. Mathematically, when fluid pressure $P$ is applied to the HFAM inner channel, it will be lengthened from the initial length $l_i$ to length $l_p$ or a displacement $x = l_p - l_i$, which is due to the circumferential constraint by the helical coil around the soft microtubule. At this displacement, HFAM accumulates EPE. When fluid pressure is removed, HFAM simultaneously discharges this EPE to generate a contraction force to an attached load and returns to its initial phase. The higher the pressure that is applied, the higher
the EPE that is stored, and thus a higher contraction force is achieved. Each HFAM has a limit of its elongated length $l_{\text{max}}$ to maintain its normal function, corresponding to maximum pressure $P_{\text{max}}$ where it achieves maximum contraction force once this pressure is withdrawn.

We here develop analytical models of generated force and motion for the HFAM with several assumptions. Firstly, the transmission fluid follows quasi-hydrostatic behavior, which means the HFAM is slowly elongated by hydraulic force. Therefore, the inertia force of the fluid is small and can be ignored. Secondly, the pressure is uniformly distributed within the inner wall of the soft microtubule. Thirdly, the helical coil constrains the radial expansion of the microtubule; thereby, the outer wall of the microtubule has no radial and circumferential deformation. Fourthly, the helical coil and microtubule elongate forward and backward with the same displacement, and the friction between them can be ignored. Lastly, the Poisson’s ratio of the microtubule is 0.5, which means its material volume is constant under deformation.

In this paper, we use the soft microtubule with an outer diameter $d_o$ that is larger than the inner diameter $d_{ic}$ of the outer helical coil (Fig. 3.4B). We leverage this advantage to overcome the initial nonlinear dead-zone of the HFAM, which is due to no change in the motion or force output regardless of an initial increase in the fluid pressure. As a result, inserting the soft microtubule into the helical coil will shrink its inner diameter to $d_i < d_o$ and stretch to a length $l_i > l_t$, where $d_i$ and $l_i$ represent the inner diameter of the soft microtubule and the initial length of the muscle at the initial phase, respectively. The outer diameter of the microtubule after insertion is equal to the inner diameter of the helical coil, $d_o = d_{ic}$. Upon pressurization by pressure $P$, the HFAM will elongate an amount of $x$ (displacement), changing its length from $l_i$ to $l_p$ and the inner diameter of the microtubule from $d_i$ to $d_p$.

Under the fifth assumption, the material volume of the microtubule is unchanged through all stages in Fig. 3.4B, we get:

$$A_t l_t = A_i l_i = A_p l_p$$  \hspace{1cm} (1)

Equation (1) is then expressed as:

$$(d_o^2 - d_{ic}^2) l_t = (d_i^2 - d_o^2) l_i = (d_o^2 - d_p^2) l_p$$  \hspace{1cm} (2)

The strain of the HFAM is defined by the ratio between its accumulated displacement and initial length $\varepsilon = x/l_i$. 

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We also define a new stretching ratio $\alpha = l_l/l_i$ for the HFAM which controls the generated force of the muscle by varying the material property in terms of its initial Young’s modulus. The value of $\alpha$ is obtained from experiments and is dependent on material properties and the original dimension of both the helical coil and the soft microtubule. Equations (1) and (2) can be rewritten as:

$$\alpha A_t = A_i = (1 + \varepsilon)A_p$$

$$\alpha (d_{ot}^2 - d_{it}^2) = d_o^2 - d_i^2 = (1 + \varepsilon)( d_o^2 - d_p^2)$$

Then, we can obtain the inner diameters of the soft microtubule when the HFAM is at its initial phase and pressurizing phase.

$$d_i = \sqrt{d_o^2 - \alpha (d_{ot}^2 - d_{it}^2)}$$

$$d_p = \sqrt{d_o^2 - \frac{\alpha}{1 + \varepsilon} (d_{ot}^2 - d_{it}^2)}$$

The generated force or contraction force of the HFAM can be described as:

$$F_{out} = F_t + F_c - F_p$$

where $F_{out}$, $F_t$, $F_c$, $F_p$ denote the exertion force of the muscle, the elastic force of the microtubule, the elastic force of the helical coil, and the driving force caused by fluid pressure, respectively.

The elastic force of the microtubule is defined by [68]:

$$F_t = \alpha EA_t \left( 1 - \frac{1}{1 + x/l_i} \right)$$

where $E$ is Young’s modulus of the soft microtubule material.

The elastic force of the helical coil obeys Hooke’s law, namely a product of the spring constant $k_c$ and displacement $x$ as follow:

$$F_c = k_c x$$

The input pressure $P$ provides the fluid force to elongate the muscle,

$$F_p = \frac{\pi}{4} d_p^2 P = P \left( \frac{\pi}{4} d_o^2 - \frac{\alpha A_t}{1 + x/l_i} \right)$$

Exertion force of the muscle in Eq. (7) can be rewritten as:
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\[ F_{\text{out}} = \alpha E A_t \left( 1 - \frac{1}{1 + x/l_i} \right) + k_c x - P \left( \frac{\pi d_o^2}{4} - \frac{\alpha A_t}{1 + x/l_i} \right) \]  \hspace{1cm} (11)

There are three distinct phases in the operating of the HFAM. In the initial phase, it is in a relaxed condition with no input pressure and no output force, \( P = 0 \) and \( F_{\text{out}} = 0 \). In the pressurizing phase, it receives input pressure to produce elongation and accumulates elastic energy, \( P = P(x) > 0 \) and \( F_{\text{out}} = 0 \). The relationship between input pressure and displacement or elongation is given in Eq. (12). In the releasing phase, it converts EPE to the contraction force. The maximum output force is obtained when the pressure is completely released, \( P = 0 \) and \( F_{\text{out}} = F_{\text{max}}(x) > 0 \). At this phase, the relationship between output force and displacement or elongation can be derived as shown in Eq. (13). There exists a reverse proportional relationship between the input pressure and the output force.

\[ P = \left[ \alpha E A_t \left( 1 - \frac{1}{1 + x/l_i} \right) + k_c x \right] \left( \frac{\pi d_o^2}{4} - \frac{\alpha A_t}{1 + x/l_i} \right)^{-1} \]  \hspace{1cm} (12)

\[ F_{\text{out}} = \alpha E A_t \left( 1 - \frac{1}{1 + x/l_i} \right) + k_c x \]  \hspace{1cm} (13)

One of the advanced features of the HFAM is that it has an additional force component \( F_c \) from the helical constraint layer, enabling it to have a higher contraction force than existing soft actuators with the same structure. In addition, the use of a new stretching ratio \( \alpha \) significantly contributes to tuning the HFAM contraction force threshold (due to an increase in its stored elastic energy) at the time it is fabricated, avoiding the use of different materials to achieve the desired output force. It means that the HFAM consists of at least three advanced components to enhance the generated force, making it a unique feature compared to the existing state-of-the-art [31, 40, 59, 63, 69].

3.3.3. Characterization of the HFAM

To characterize the performance of various aspects of a single HFAM, we built a measuring system that can analogously collect the output elongation and contraction force corresponding to input hydraulic volume and pressure (Fig. 3.5). The system consists of a DC-motor, amplifier, pressure sensor, miniature syringe, linear ball screw, encoder, load cell, and real-time controller. The actuation unit was equipped with a DC-motor coupling with a ball screw to provide linear motion to the syringe plunger. An encoder was placed at the motor side to track the displacement of the plunger, then be converted to the input volume. The input pressure was collected by a pressure sensor located amid the fluid transmission tube by a T-connector. One end of the HFAM
was connected to the syringe via the fluid transmission tube, while the other end was connected to either an encoder to collect the accumulated displacement signal or a load cell to obtain force data. A thin, long, and highly stretchable elastic string was used to retain the HFAM from slack while collecting the elongation signal. This system can generate input pressures of up to 2 MPa to the HFAM. The experimental platform was designed as a modular system where the HFAM can be easily exchangeable. Detailed specifications of the components used in experiments are available in Table 3.4, Appendix.

**Figure 3.5.** Experimental setup for the characterization of HFAM (both elongation and force). (A) Schematic diagram. (B) Actual experimental setup (upper panel-elongation test and lower panel-force test).
**Elongation and force characterization**

We used an 80 mm long muscle embodiment HFAM1 (detailed specifications are shown in Table 3.1) in the experiments. We applied 0.1 Hz sinusoidal signals to the syringe plunger, which controls the fluid volume and pressure to the HFAM1 and constructed the relationship between the input (volume and pressure) and output (elongation and force). Experimental results for elongation and force response are given in Fig. 3.6. We observed that the 80 mm long HFAM1 could reach approximately 81.4% elongation with 0.125 ml of supplied fluid volume, which corresponds to 1.32 MPa in maximum pressure. Experimental results also revealed that the hysteresis profiles in forward and backward directions showed noticeable distinctions. In the volume-elongation chart (Fig. 3.6A), the pressurizing phase profile almost overlaps the releasing phase profile. This identical occurrence due to the incompressible property of fluidic means (water). Every volume change was well captured by the muscle elongation regardless of which phases it was engaging. In contrast, the pressure-elongation plot (Fig. 3.6B) has a relatively wide hysteresis profile, in which a larger gap exists between the pressurizing and releasing phases. This behavior reflects the energy loss of the muscle when changing between the two working phases. It discharges a smaller amount of energy in the releasing phase than it receives in the pressurizing phase. For the contraction force response, the hysteresis profile has a reverse proportional relationship with respect to the change of input volume and pressure. When pressurizing to 86.3% of elongation, the HFAM1 could generate a force of 1.08 N. There are relatively narrow gaps in the hysteresis profiles of volume-force (Fig. 3.6C) and pressure-force (Fig. 3.6D). We observed a nearly linear relationship between force and pressure at pressurizing and releasing profiles in Fig. 3.6D. Unlike the elongation test where the muscle was being switched between stressed and relaxed states, the muscle in the force experiment was constantly in the elastic-storage mode where it was highly sensitive with any input variations.

To better understand the HFAM performance in terms of stored elastic energy or elongation, we also performed another experiment to establish the relationship between elongation and output contraction force compared with our developed models. We examined three different muscle lengths (30 mm, 50 mm, and 80 mm) of embodiment HFAM1 (Table 3.1). We gradually increased the elongation from 0% to 150% with a step of 10%. The maximum force values were collected at each blocked elongation; data mean and standard deviations of five testing cycles are presented in Fig. 3.7A. The HFAM contraction force has a proportional relation with the elongation, on which there is a nearly linear profile until 100% strain, following by an exponential increase. All three different muscle lengths share the same trajectory with minor
deviation. When the initial elongation of the HFAM increases, it accumulates more elastic energy, leading to a stronger contraction force when releasing the pressure. The dashed line represents the HFAM analytical model following Eq. (13). Input parameters were obtained from either manufacturer specifications or experiments and are as follows: \( \alpha = 0.9, \quad E = 1.648 \text{ N/mm}^2 \) (at 100\% strain), \( A_t = 0.8026 \text{ mm}^2, \quad l_{i30} = 30 \text{ mm}, \quad k_{c30} = 0.0183 \text{ N/mm}, \quad l_{i50} = 50 \text{ mm}, \quad k_{c50} = 0.011 \text{ N/mm}, \quad l_{i80} = 80 \text{ mm}, \quad k_{c80} = 0.0069 \text{ N/mm} \). Our developed model neatly followed experimental data up to 100\% elongation but was unable to capture the HFAM behavior at higher strain. The underlying mechanism of this phenomenon is the augmentation of the microtubule Young’s modulus at high strain while our analytical model supposed it as a constant variable at 100\% strain.

**Figure 3.6.** Elongation and force responses of a single HFAM (1.49 mm in diameter, length 80 mm) with respect to the change of applied input volume and input pressure. (A, B) Relation between output elongation and applied hydraulic volume and pressure, respectively. (C, D) Hysteresis profiles of the relation between output force and applied hydraulic volume and pressure, respectively.
Table 3.1. HFAMs used in experiments

<table>
<thead>
<tr>
<th>Variant</th>
<th>Helical coil</th>
<th>Microtubule</th>
</tr>
</thead>
<tbody>
<tr>
<td>HFAM1</td>
<td>Stainless steel (Asahi Intecc Co., Ltd., Japan)</td>
<td>Silicone rubber (Saint-Gobain S.A., France)</td>
</tr>
<tr>
<td></td>
<td>OD: 1.49 mm</td>
<td>OD: 1.19 mm</td>
</tr>
<tr>
<td></td>
<td>ID: 1.10 mm</td>
<td>ID: 0.64 mm</td>
</tr>
<tr>
<td></td>
<td>$k_c$: 0.011 N/mm (length 50 mm)</td>
<td>E (100%): 1.648 MPa</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Hardness: Durometer 53A</td>
</tr>
<tr>
<td>HFAM2</td>
<td>Stainless steel (McMaster-Carr Supply Co., USA)</td>
<td>Latex rubber (McMaster-Carr Supply Co., USA)</td>
</tr>
<tr>
<td></td>
<td>OD: 3.18 mm</td>
<td>OD: 3.18 mm</td>
</tr>
<tr>
<td></td>
<td>ID: 2.51 mm</td>
<td>ID: 1.59 mm</td>
</tr>
<tr>
<td></td>
<td>$k_c$: 0.068 N/mm (length 100 mm)</td>
<td>E (100%): 1.856 MPa</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Hardness: Durometer 40A</td>
</tr>
</tbody>
</table>

$k_c$: spring constant; E (100%): Young’s modulus at 100% strain

**Frequency response**

We evaluated the HFAM response to oscillating inputs with various frequencies. We used the setup similar to those in Fig. 3.5 where a motorized linear slider (Zaber, model X-LRQ150BL-E01, Zaber, Canada) provides input volume to a syringe driving a 50 mm long embodiment HFAM2. The Zaber generated a sinusoidal chirp signal that has a fixed amplitude of 5% of the specimen elongation and incremental frequencies from 0.2 Hz to 20 Hz. We used a laser sensor (Keyence, model IL-100, Keyence Corp., USA) to monitor the displacement of the HFAM distal tip. We collected this displacement for five testing cycles, then calculated its mean and standard deviation at each frequency. Subsequently, to quantify amplitude loss, we reflected the amplitude reduction of each frequency to the first one by the root-power quantity equation $M(f) = 20 \log(T_f/T_0)$, where $M(f)$ and $T_f$ denotes signal magnitude (dB) and amplitude (mm) at frequency $f$, respectively; $T_0$ is the amplitude of the first frequency (0.2 Hz). Experimental result (Fig. 3.7B) showed a plunge of signal magnitude until 3 Hz (-0.95 dB), followed by a steady decrease of about -0.039 dB/Hz for the remaining frequencies. At 10 Hz and 20 Hz, the HFAM has lost its amplitude of 12.4% (-1.15 dB) and 16.9% (-1.61 dB), respectively. The result also revealed that our HFAM responded well to high-frequency input signals, potentially benefitting in applications.
that require high operating frequencies. We were unable to indicate the typical -3 dB cut-off frequency because of mechanical system limitations which could not generate input signals with frequency larger than 20 Hz.

Figure 3.7. Analytical model validation and frequency response. (A) Model validation using embodiment HFAM1 with three different lengths (30 mm, 50 mm, and 80 mm). (B) HFAM frequency response using a 50 mm long embodiment HFAM2 running at 5% of elongation.

*Durability and elongation limit*

We conducted durability tests to demonstrate the reliability and reusability of the HFAM. An embodiment (HFAM1, Table 3.1) with an original length of 80 mm was examined through
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elongation and force durability tests. The experimental setup was similar to those in the previous section. We applied sinusoidal signals to the DC-motor at 0.1 Hz over 2000 seconds. Results (Fig. 3.8A and Fig. 3.8B) revealed that both elongation and force data almost remained stable to the input signals. However, the signal amplitudes are gradually shrinking over time. The force signal exhibits a larger shrinking rate compared to the elongation signal. After 2000 seconds, the force shrinks 11.9% from 0.957 N to 0.843 N, while elongation shrinks only 1.9% (elongation reduces from 81.36% to 79.82%). This reduction of amplitude over time can be explained by the inherent elastic properties of the HFAM. The outer coil and microtubule are elastic components affected by fatigue after performing multiple periodic cycles of stretching. A huge difference in the shrinking rate of elongation and force is mainly because of the nature of the experimental setup. The HFAM in the elongation test was being stretched in the pressurizing phase and then relaxed. In contrast, the force test maintained a constant elongation of the HFAM while the data collection process was running. As a result, the accumulated fatigue of the HFAM was augmented in the case of the force test in comparison to the elongation test.

We also examined the maximum elongation of the HFAM using a 50 mm long HFAM1. The muscle was being switched between the initial phase and pressurizing phase, with a steady increase of signal amplitude at each cycle. In the last full cycle, the HFAM1 achieved an elongation of 224% and successfully returned to its initial phase. Subsequently, the HFAM reached its elongation limit of 246.8% before malfunctioning (Fig. 3.8C). We performed ten trials to validate our experiment (246.8% ± 8%) and observed that most of the failure (9 out of 10) originates from the interface between the fluid transmission tube and the soft microtubule. It means that if a stronger adhesive glue is used, our HFAM can reach a greater elongation limit. We applied the same process for embodiment HFAM2 and received the maximum elongation of 102% ± 6% for ten trials.

*Lifting performance and energy efficiency*

We conducted experiments to characterize the lifting performance and energy efficiency of the HFAM. To demonstrate the scalability, we used two different HFAMs (each had an original length of 50 mm, details are shown in Table 3.1). We performed lifting tests in which the HFAMs were connected to their relevant weights (Fig. 3.9A). We supplied input pressure to the HFAMs to store elastic energy while the weights were simultaneously lowered. We then withdrew the fluid pressure to induce the contraction force to lift the weight up. Results showed that the single 50 mm long HFAM1 that weighed 0.28 g could lift a load of 100 g, which was about 357 times
larger than its weight, with a stroke of 64 mm (95.5% of elongation). Likewise, the single 50 mm long HFAM2 could lift a load that was 352 times heavier than its mass (500 g compared to 1.42 g) but achieved a lower stroke, at 42 mm.

Figure 3.8. HFAM durability and elongation limit. (A, B) Durability tests of a single HFAM for elongation and force. The left enlarged panel represents the first 50 seconds while the right enlarged panel shows the last 50 seconds. (C) Maximum elongation of embodiment HFAM1 and HFAM2.

We also calculated the energy efficiency of the previous HFAM variants based on the lifting performance data. Energy efficiency reflects how effective the HFAM is in converting energy consumption to mechanical power. It is defined by the ratio between the output and input mechanical power. In the experiments, HFAMs were powered by fluid (water) via a standard 1 ml syringe. Therefore, the input mechanical power is a product of the applied force $F_{pla}$ to the
plunger and displacement of the plunger $x_{plu}$ inside the syringe barrel per moving time $t_{in}$. The applied force was measured by a force gauge (MARK-10, USA) which was directly connected to the syringe plunger. The output mechanical power is a product of a lifting load $F_{load}$ and moving distance (stroke) $x_{load}$, per lifting time $t_{out}$. However, the instant response of the HFAM makes an insignificant deviation between $t_{in}$ and $t_{out}$ due to low nonlinear hysteresis. In this test, we try to understand the performance of the HFAM with respect to different sizes and strokes. Therefore, we assumed that this deviation is small and can be ignored or $t_{in} \approx t_{out}$. Experimental data of the two HFAM variants and their energy efficiency $EE = F_{load}x_{load}/F_{plu}x_{plu}$ are calculated and presented in Table 3.2. Experimental results revealed that the single 50 mm long HFAM1 and HFAM2 could reach an energy efficiency of 46% and 62.7%, respectively. Despite more energy consumption, the key point of the HFAM2 to surpass the HFAM1 in terms of energy efficiency is the ability to carry a much higher load.

**Figure 3.9.** Lifting performance and the ability to pre-stored elastic energy. (A) Embodiment HFAM1 and HFAM2 lift weights of 100 g and 500 g, respectively. (B) HFAM with a smaller stretch ratio $\alpha$ exerted a shorter extension when holding the same load.

*Tunable generated force via stretch ratio*

We also performed experiments to demonstrate our HFAMs can tune their elastic energy to increase generated force via the change of the stretch ratio $\alpha$ of the inner microtube. This is challenging to obtain in current fluid-driven SAMs. Experimental results (Fig. 3.9B) showed
the HFAM generated force could be tuned by adjusting its stretch ratio \( \alpha \) at the time it was fabricated, enabling an increase in its output force. To demonstrate this advanced feature, an HFAM with a small stretch ratio \( \alpha = 0.63 \) (muscle length 35 mm, microtubule length 22 mm) was fabricated. Due to an increase in its Young’s modulus, this HFAM exhibited stronger elastic energy, resulting in a stronger holding force, which was evidenced by a smaller elongation of the muscle against an external load. In contrast, the HFAM with a higher stretch ratio \( \alpha = 1 \) (muscle length 35 mm, microtubule length 35 mm) exhibited weaker elastic energy due to no change in its original Young’s modulus, resulting in smaller holding force or a higher elongation of the muscle against the external load.

<table>
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<tr>
<th>Variant</th>
<th>Input power</th>
<th>Output power</th>
<th>Energy efficiency</th>
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<tr>
<td></td>
<td>Force (N)</td>
<td>Displacement (mm)</td>
<td>Load (g)</td>
</tr>
<tr>
<td>HFAM1</td>
<td>23.8</td>
<td>5.73</td>
<td>100</td>
</tr>
<tr>
<td>HFAM2</td>
<td>21.25</td>
<td>15.47</td>
<td>500</td>
</tr>
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</table>

3.4. Performance of HFAM-like tendon-driven mechanisms

Tendon-driven mechanisms have been widely used in many fields, ranging from surgical robots and medical devices, wearable haptic systems, soft assistive devices for rehabilitation and human augmentation, space missions, and others [2, 22]. Despite advances, these mechanisms have intrinsic drawbacks of high force loss due to highly nonlinear hysteresis and backlash associated with a long, complex, and tortuous path, leading to a complex control system and degradation of energy efficiency [67]. The HFAM-like tendon-driven mechanisms developed here are expected to revolutionize these prolific fields by providing a more effective yet simple soft miniature actuator, offering additional options in the development of emerging medical devices and robotic systems. In this section, we will demonstrate several configurations of the HFAM to simulate tendon-driven mechanisms, including an HFAM-like tendon-pulley mechanism and an HFAM-like tendon-sheath mechanism.
Figure 3.10. Accumulative bending angle and two forms of HFAM-like tendon-driven mechanisms. (A) Illustration of accumulative bending angle. (B, C) Experimental setup for the HFAM-like tendon-pulley mechanism and HFAM-like tendon-sheath mechanism with (1) actuation unit, (2) load cell, (3) syringe and fluid transmission tube, (4) HFAM, (5) pulley, (6)
load, (7) laser sensor, and (8) outer tube (sheath). (D) Performance of the HFAM-like tendon-pulley mechanism with different configurations. (E) Performance of the HFAM-like tendon-sheath mechanism with different configurations. (F) Another configuration of the HFAM-like tendon-pulley mechanism (the HFAM is a type of HFAM2, Table 3.1) with mechanical pins serving as routing pulleys. (G) Another configuration of the HFAM-like tendon-sheath mechanism (the HFAM is a type of HFAM2, Table 3.1) where the HFAM served as a tendon and the outer tube served as a routing sheath.

3.4.1. HFAM-like tendon-pulley mechanism

Here we demonstrate an HFAM configured as a tendon-pulley mechanism to assess its energy efficiency. In applications with a tortuous path or multiple loops, force loss of the tendon-pulley mechanism mainly originates from bending shapes where the path directions change. The accumulative angle, which was detailed in [24], is typically used to describe the configuration of a flexible actuator and is defined by the total bending angles of n bending segments throughout the path \( \theta = \sum \theta_i (i = 1 \sim n) \) (Fig. 3.10A). To demonstrate that our HFAM offers better performance compared to the conventional tendon-driven mechanism under varied accumulative angles, we fabricated a 400 mm long embodiment HFAM2 (Table 3.1) to perform lifting tests under various configurations of the transmission path. The experimental setup is shown in Fig. 3.10B, where an actuation unit (Zaber, Canada) generated input displacement to the syringe plunger with a load cell in between to measure the input force. In this experiment, the HFAM2 was fixed at one end and its other end was connected to a load of 200 g. A laser sensor was used to measure the weight displacement. For illustration, the HFAM2 was configured to three different transmission paths where the HFAM body was routed via a set of pulleys with the corresponding accumulative angle \( \theta \) of \( \pi \), \( 1.5\pi \), and \( 2\pi \). We applied 10 mm displacement to the syringe plunger and simultaneously measured input force and output weight displacement. Results (Fig. 3.10D) revealed that all three configurations achieved approximately a similar energy efficiency at 29%. There are slight decreases in lifting performance when the accumulative angles increase (energy efficiency reduced by 0.33% and 0.75% for the \( 1.5\pi \) and \( 2\pi \), respectively, compared to the \( \pi \) configuration). These results confirmed that our HFAM-like tendon pulley mechanism has a small change of energy efficiency with respect to different accumulative angles. In addition, we also noted that this energy efficiency remained almost the same regardless of the length of the fluid transmission tube.
3.4.2. HFAM-like tendon-sheath mechanism

We also studied the effectiveness of the HFAM as a tendon-sheath mechanism with various configurations. In surgical and wearable applications where the long and dexterous paths from the power source and the distal actuator are complex, the use of a tendon-sheath mechanism is highly desired. We conducted lifting tests and evaluated the energy efficiency of the HFAM-like tendon-sheath mechanism with the experimental setup (Fig. 3.10C) similar to that of the previous section. We used a 100 mm long embodiment HFAM1 (Table 3.1) which was routed inside a plastic hollow tube (inner diameter of 2.85 mm, length of 2 m). The outer tube acted as an outer sheath that could assume different configurations. As an illustration, we used three different configurations with the accumulative angles of $3.5\pi$, $5.5\pi$, and $7.5\pi$, respectively. These multiple-loop configurations happened at the fluid transmission tube section where the shape at the muscle section was changed very little or remained unchanged in order to simulate a routing path similar to that of a conventional tendon-sheath mechanism. We input 2 mm displacement at the syringe plunger to lift a load of 50 g. Experimental results (Fig. 3.10E) showed insignificant distinctions between the three configurations. The $3.5\pi$ and $5.5\pi$ both achieved about 83% energy efficiency while this number slightly decreased in the $7.5\pi$, to 82%. This variation can be explained by the experimental error during the setup, which is due to existing microbubbles within the fluid transmission tube.

It is clear that the tortuous path with multiple loops of the fluid transmission tube creates a minor effect on muscle performance, leading to a similar energy loss compared to conventional tendon-sheath mechanism in flexible robotic applications such as surgical instruments or soft exoskeleton. These results demonstrated that our HFAM offers excellent performance regardless of the varied routing configuration or the change of accumulative angle. This means that the developed HFAM can be a potential actuation candidate for applications that require complex and long routing paths between the external power driving source and the distal end of the actuator. In addition, the little or no change of the energy efficiency indicates that our HFAM is well-suited for feedforward-based compensation for nonlinear hysteresis where no real-time output feedback is required during the compensation. This makes the HFAM a well-suited candidate for flexible surgical robots or soft wearable devices where the integration of onboard sensors is challenging. To further demonstrate the usability of HFAM-like tendon-driven mechanisms, we also carried out experiments to evaluate the HFAMs with different types of routing components including mechanism pins simulated as routing pulleys (Fig. 3.10F) and with a silicone tube that served as a routing sheath (Fig. 3.10G).
3.5. HFAM-driven soft wearable robotic glove

Here we demonstrate the use of the HFAM-like tendon-driven mechanisms for the development of a soft wearable glove to potentially assist the wearers with hand impairment or to augment their movement. There is a noticeable paradigm shift from heavy, bulky rigid-frame exoskeletons to soft, wearable counterparts due to the facile and compliant human-robot interaction. An HFAM-based soft robotic glove is created to actively assist the human hand’s grasping motion with generated bending motion of the wearer’s fingers (thumb, index, and middle) simultaneously. Most soft wearable gloves use tendon-sheath mechanisms to convey actuation force from a remote driving source. However, the use of tendon-sheath mechanisms with long routing paths introduces high force loss and nonlinear hysteresis, as discussed in previous sections. We leverage the advantage of low friction loss HFAMs to fabricate a soft wearable glove where its driving source is multiple HFAM2s (Table 3.1) that are routed along the forearm to power the fabric glove fingers via short routing tendon-sheath mechanisms. We used short routing tendon sheaths to avoid a high friction loss while maintaining a compact wearable form factor to minimize any potential discomfort to the wearer. The new glove was designed in a way that is easy to wear and able to perform several grasping tasks with various objects of different shapes and sizes.

3.5.1. Design, fabrication, and working principle

The soft exoskeleton glove was developed using a simple and effective fabrication approach where all required components (tendons and outer sheaths) were directly integrated into a commercial fabric glove (Fig. 3.11A). First, conventional tendons were routed alongside the fingers at both the palm and back sides of the hand via miniature routing nodes and short routing sheaths. It is noted that the backside sheaths were continuous while the palm side sheaths appeared in hollow nodes to avoid obstacles and maximize the grasping range. The nodes were used to guide flexible tendons to embrace the fingertips firmly without using additional thimbles. A smooth and continuous junction between the index and middle finger enabled the wearer’s fingers to be controlled at the same time as a reduction in the number of required actuators. All routing tendon-sheaths were converged to the hand’s backside where they were anchored into a wrist strap. This arrangement allows gathering the actuation mechanism in the backside, enabling free space in the palm side to facilitate grasping gestures. The glove was driven by four HFAM2s, in which the first two HFAM2s simultaneously controlled the flexion and extension of both the index and middle finger while the last two HFAM2s manipulated the thumb. To stabilize the
muscle during the operation, the proximal ends of the HFAM2s were anchored by another strap located near the elbow. Four flexible coils serving as routing sheaths were placed between two straps to guide the HFAMs.

Figure 3.11. Overview of the soft exoskeleton glove with its components. (A) Design model with routing tendon-sheath and HFAMs-like tendon sheath mechanisms. (B) Kinematic diagram in the x-y plane of a single finger driven by both flexion and extension tendons via miniature routing nodes.
A simple kinematic diagram in the $x$-$y$ plane (Fig. 3.11B) describes the methodology to control the bi-directional motion of a finger. The wearer’s palm plays the role of a static base, enabling bending motion of the fingers from the driving tendons. The routing nodes are distributed in the middle of the finger phalanges where they act as anchor points to receive force transmitted from the tendons. There are two flexion routing tendons placed in off-centered positions at the palm side for each finger. This alignment stabilizes the direction of flexion force when grasping and eliminates the tendons’ interference with their routing nodes and objects. In contrast, the extension motion only requires a single tendon at the finger centerline. Antagonistic motions of flexion and extension tendons, driven by HFAM2s, manipulate the corresponding motions of the controlled fingers. The HFAM2s are driven by DC-motor actuation units via syringes, linear ball screws, and fluid transmission tubes.

3.5.2. Prototype and grasping performance

We evaluated the performance of a soft exoskeleton glove in free space and with objects. We used a nylon glove with polyurethane coating placed on the palm and fingers (Avit AV13074, Carl Kammerling International Ltd, UK) as an outer skin for the glove. We used stainless steel helical coils (inner diameter of approximate 5.4 mm and outer diameter of 6.35 mm, McMaster Carr, USA) as routing sheaths for the HFAM2s and stainless-steel tendons with Teflon coating (diameter of 0.5 mm, Asahi Intec Co., Ltd., Japan) as the driving tendons. The routing sheath for the driving tendons were stainless steel coils (inner diameter of 1.1 mm and outer diameter of 1.49 mm, Asahi Intec Co., Ltd., Japan). The hollow nodes, 4 mm in length, were cut from a stainless-steel coil (inner diameter of 1.1 mm and outer diameter of 1.49 mm, Asahi Intec Co., Ltd., Japan). All routing sheaths and nodes for the driving tendons were attached to the glove by sewing thread and reinforced by super glue. The wrist strap and forearm strap were made from polylactic acid (PLA+, Shenzhen Esun Industrial Co., Ltd., Shenzhen, China) by a 3D printer (Ultimaker 2+, Ultimaker B.V. Utrecht, Netherlands), then attached into velcro straps as convenient fasteners (Fig. 3.12A). After investigating the distance between two straps and required strokes of each HFAM to fully assist the grasping motion, we fabricated four HFAM2s (Table 3.1) with the corresponding lengths presented in Table 3.3. With these given lengths, HFAMs can have a pre-tension at about 5% - 10% strain and achieve the desired strokes at 65% of their elongation.
Figure 3.12. Final prototype of the HFAMs-driven soft fabric glove. (A) Overview of components of the soft glove including a soft fabric glove, driving hydraulic source (DC-motors, linear ball screws, and miniature syringes), HFAM-like tendon-sheath mechanisms, and short conventional tendon-sheath mechanisms. (B) Grasping gestures of the soft glove in free space without objects.

Table 3.3. HFAMs used in experiments

<table>
<thead>
<tr>
<th>Task</th>
<th>Required stroke (mm)</th>
<th>HFAM length (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Index and middle finger</td>
<td>Flexion 65</td>
<td>130</td>
</tr>
<tr>
<td></td>
<td>Extension 30</td>
<td>60</td>
</tr>
<tr>
<td>Thumb</td>
<td>Flexion 32</td>
<td>60</td>
</tr>
<tr>
<td></td>
<td>Extension 30</td>
<td>60</td>
</tr>
</tbody>
</table>

We demonstrate the grasping performance of the glove in free space without objects and with various objects of different sizes. A human subject (user) with healthy hands wore the glove and performed grasping tasks when all fingers were relaxed. The HFAM2s entirely controlled the finger motions through the DC-motor actuation units. We applied a simple open-loop
control scheme to drive the robotic glove. The required stroke of each muscle (Table 3.3) was mapped to the input volume of its corresponding syringe driven by its designated actuation unit. To execute bi-directional motions of a finger, the actuation unit for flexion motion has been assigned inverse direction with its extension counterpart. The experiments without objects were carried out to examine the user’s comfort with different grasping gestures (Fig. 3.12B). Subsequently, we performed grasping experiments with objects and results showed that the soft wearable glove could assist a variety of grasping tasks with arbitrary surfaces such as an apple, lemon, a weight of 500 g, and a glass beaker (Fig. 3.13). The glove could also support a firm grasp that allowed the subject to lift and move the object without slippage. Both grasping and releasing motions showed a relatively fast response. Real-time performance of the glove is also given in the Supporting Video.

Figure 3.13. Prototype and performance of the soft exoskeleton glove. (A) Backside view of the prototype with HFAMs being covered inside stainless-steel coils. (B) Grasping demonstration with multiple objects: (1) Apple; (2) Lemon; (3) Weight 500 g; (4) Glass beaker.
3.6. Discussion, conclusion, and future work

A soft hydraulic filament artificial muscle (HFAM) that can be simply fabricated using self-made or off-the-shelf materials at scale and well suited for mass production has been developed. The simplicity of the insertion method enables the new HFAM to achieve miniature sizes and high aspect ratio. The HFAM, controlled by fluid pressure, can elongate to store elastic energy and then exert a contraction force when depressurizing. The higher the pressure supplied to the HFAM, the more it elongates and accumulates energy, resulting in a higher contraction force when this energy is discharged. Both elastic components (microtubule and helical coil) of the HFAM actively contribute to enhancing its contraction force. Also, the HFAM offers the force-tuning capability by manipulating the stretch ratio.

Our HFAM reached 0.15 MPa of stress, 246.8% of strain, and 62.7% of energy efficiency. It outperformed a typical mammalian skeletal muscle with typical stress at 0.1 MPa, maximum strain at 40%, and maximum efficiency of 40% [70]. We also demonstrated the HFAM capability as a tendon-driven mechanism. Compared to conventional tendon-driven mechanisms [2, 22, 67], our HFAMs offer smaller energy loss regardless of long and tortuous transmission paths. This outstanding property may revolutionize the field of surgical robotics, medical instruments, and flexible assistive devices. The developed HFAMs were implemented to a soft exoskeleton glove that can provide useful assistance to the wearers to grasp and hold multiple objects.

Despite advances, the HFAM required relatively high pressure to operate; for example, it needed around 1.3 MPa to generate 80% elongation. Also, the current HFAM lacks an onboard sensing system to provide real-time feedback for nonlinear hysteresis identification and control. Therefore, it is recommended that future work should be focused on exploring new silicone materials with higher strain and stress that can provide self-sensing capabilities for feedback control. Advanced nonlinear hysteresis models (based on Bouc-Wen or Prandtl-Ishlinskii models [71, 72]) that can capture the HFAM behavior at high strain are desirable.

Furthermore, to enhance the actuation accuracy, it is necessary to investigate advanced hydraulic system controls such as extended-state-observer-based adaptive control [73], RISE-based adaptive control [74, 75]. Upon achieving sensing ability by either an embedded microtubule sensor or self-sensing, the feedback signals will be fed to the adaptive controller for real-time compensation with disturbances and uncertainties.
Chapter 3. Hydraulic filament artificial muscles

In conclusion, we have introduced a new generation of soft artificial muscles made from a helical coil and a microtubule suitable for mass production. The developed HFAMs are scalable and have a high aspect ratio. The filament muscles achieved high elongation, fast response, and high energy efficiency. We believe that the developed muscles with demonstrated applications would significantly contribute to soft robotic research and commercial products.

References


## Appendix

**Table 3.4.** Detailed specifications of experimental components

<table>
<thead>
<tr>
<th>Component</th>
<th>Model</th>
<th>Specifications</th>
<th>Manufacturer</th>
</tr>
</thead>
<tbody>
<tr>
<td>DC-motor</td>
<td>3272G024CR</td>
<td>Gearhead: 68:1</td>
<td>Faulhaber, Germany</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Encoder: IE3-1024</td>
<td></td>
</tr>
<tr>
<td>Ball screw</td>
<td>SFU1204</td>
<td>Shaft: 12 mm</td>
<td>XINHUANGDUO AUTO, China</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Lead: 4 mm</td>
<td></td>
</tr>
<tr>
<td>Syringe</td>
<td>Luer-Lok™ 1 mL</td>
<td>Volume: 1 ml</td>
<td>BD Biosciences, Canada</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Ratio: 57.3 mm/ml</td>
<td></td>
</tr>
<tr>
<td>Pressure sensor</td>
<td>40PC250G2A</td>
<td>Capacity: 250 psi</td>
<td>Honeywell, USA</td>
</tr>
<tr>
<td>Encoder</td>
<td>S6S-1000-B</td>
<td>Optical, 1000 CPR</td>
<td>US Digital, USA</td>
</tr>
<tr>
<td>Load cell</td>
<td>LSB200</td>
<td>Capacity: 10 lb</td>
<td>FUTEK, USA</td>
</tr>
<tr>
<td>Controller</td>
<td>QPIDe</td>
<td>8 channels</td>
<td>Quanser, Canada</td>
</tr>
<tr>
<td>Force gauge</td>
<td>MARK-10, Series 5</td>
<td>Capacity: 25 N</td>
<td>MARK-10, USA</td>
</tr>
<tr>
<td>Motorized linear slider</td>
<td>X-LRQ150BL-E01</td>
<td>Stroke: 150 mm</td>
<td>Zaber, Canada</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Resolution: 0.5 µm</td>
<td></td>
</tr>
<tr>
<td>Laser sensor</td>
<td>IL-100</td>
<td>Range: 100±20 mm</td>
<td>Keyence, USA</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Accuracy: ±0.05%</td>
<td></td>
</tr>
</tbody>
</table>
Chapter 3 investigated the fundamental development of a soft hydraulic filament artificial muscle including design, fabrication, and performance characterisation. Chapter 3 also presented the first application of the new artificial muscle as an actuation mechanism for a soft wearable robotic glove. This chapter further exploits the proposed artificial muscle in its single filament configuration for smart surgical sutures for wound closure. The contraction motion of the artificial muscle upon depressurisation is utilised to approximate two tissue margins. Smart surgical sutures can be knotted like conventional surgical sutures or equipped with newly developed anchoring features to secure tissues. A new mathematical model that accounts for the dynamic Young’s modulus of the silicone tube is developed to describe the elongation-force relationship of the proposed artificial muscle. Three types of anchors are introduced and their puncturing and holding force through real porcine stomach tissue is experimentally tested. The use of smart surgical sutures is demonstrated in ex-vivo experiments for common procedures such as perforation closure and tissue folding (weight loss surgery) on porcine stomach and colon tissue.

This work has been published:

4.1. Abstract

Wound closure with surgical sutures is a critical challenge for flexible endoscopic surgeries. Substantial efforts have been introduced to develop functional and smart surgical sutures to either monitor wound conditions or ease the complexity of knot tying. Although research interests in smart sutures by soft robotic technologies have emerged for years, it is challenging to develop a soft robotic structure that possesses a similar physical structure as conventional sutures while offering a self-tightening knot or anchor to close the wound. This paper introduces a new concept of smart sutures that can be programmed to achieve desired and uniform tension distribution while offering self-tightening knots or automatically deploying secured anchors. The core technology is a soft hydraulic artificial muscle that can be elongated and contracted under applied fluid pressure. Each suture is equipped with a pressure locking mechanism to hold its temporary elongated state and to induce self-shrinking ability. The puncturing and holding force for the smart sutures with anchors are examined. Ex-vivo experiments on fresh porcine stomach and colon demonstrate the usefulness of the new smart sutures. The new approaches are expected to pave the way for the further development of smart sutures that will benefit research, training, and commercialization in the surgical field.

4.2. Introduction

Wound closure is a method to accelerate the healing process of defect tissue at the surgical site [1]. The use of surgical sutures is the most common practice in the medical field as it can promote faster wound healing by connecting tissue with small scar formation. To achieve precise wound closure for the healing process, the ideal suture should possess sufficient mechanical strength and uniform tension distribution as well as high flexibility to enable less suturing complexity during operation [2, 3]. However, none of the current commercial sutures can satisfy all these requirements. A variety of materials such as silver, gold, steel wires, animal hair, silk, plant fibers, or tree bark have been used for surgical sutures in the last 4000 years [4]. Recent advances in functional materials have witnessed the use of synthetic biomaterials such as poly (lactic-co-glycolic acid) or polydioxanone as base materials for surgical sutures. Despite advances, no unique material can meet all types of surgical applications [3-6]. The choice of suture materials for wound closure is closely related to several factors such as layers of tissues, tension, tissue type, surgical access, and time of removal. The use of absorbable and nonabsorbable materials for sutures also depends on the type of wound closure [3, 4, 7].
Absorbable sutures normally undergo degradation by losing around 50% of their tensile strength within 60 days in the tissues where they are either digested by enzyme or hydrolyzed while nonabsorbable ones exhibit a poor degradation in the body tissues. Some nonabsorbable sutures such as steel sutures for sternal closure or bone can be left permanently in the body.

Advances in minimally invasive surgery such as flexible endoscopy have gained popularity over the past few years. Flexible endoscopic surgeries offer many benefits compared with conventional abdominal open surgery and invasive laparoscopic surgery, including less postoperative pain, faster recovery, and no abdominal skin scars [8]. However, risks of perforation, both accidentally and purposely, have remained a critical challenge for endoscopic procedures [9, 10]. Endoscopic submucosal dissection (ESD) is used to remove tumors in the gastrointestinal mucosal layer. ESD has a high risk of perforation (5–10%) [11] and so requires surgeons with exceptional skills. Endoscopic full-thickness resection (EFTR) is a new method to deal with overgrown tumors in which en bloc tumor removal penetrates the gastrointestinal muscular layer, causing a perforation [11]. Natural orifice transluminal endoscopic surgery (NOTES) is an advanced endoscopic procedure for minimally invasive abdominal surgery where an internal incision in the stomach, colon, or bladder is required for instruments to access the abdominal cavity [9, 12]. This internal full-thickness perforation is purposely created and requires closure after finishing the targeted procedure. One of the biggest challenges for the use of surgical sutures for both laparoscopy and endoscopy is the difficulty to close defect tissue effectively with precise sewing and desired tension while maintaining strong knot tying at the surgical site [13, 14]. If the suture tension is too high, necrosis of the enclosed and surrounding tissue may occur, leading to unrelieved pain and other unexpected complications for the patients. If the suture tension is too weak, scar tissues with poor mechanical properties may occur, leading to potential hernia and chronic non-healing wounds [3, 4].

Perforation closure is vital to advance the development and enable the full potential of flexible endoscopic surgery, making it an important paradigm shift for minimally invasive endoscopic surgery. Recognizing this prospect, research on an effective solution for perforation closure has proliferated, with a primary focus on the improvement of both surgical sutures and closure devices [15, 16]. The research direction can be divided into three main categories. The first approach is the use of conventional surgical sutures and combines them with new endoscopic suturing devices. For example, the OverStitch endoscopic suturing system (Apollo Endosurgery, Inc., USA) offers full-thickness sutures for both running and interrupted stitches through a curved needle [17]. The needle punctures and drives the suture through tissue, then
two T-tags are deployed to secure closure without the need to tie complex surgical knots. The incisionless operating platform (IOP) (USGI Medical, Inc., USA) has been used for various endoscopic suturing procedures. The platform has multiple tools that can capture and puncture to pass a suture through tissue, then deploy a pair of mesh-shaped expandable tissue anchors to deliver a secure closure [18]. Besides gastrointestinal perforation closure, OverStitch has been used for endoscopic sleeve gastroplasty (ESG) and the IOP used for primary obesity surgery endoluminal (POSE). While weight loss treatments can be performed via capsule endoscopy [19, 20], in general these surgical procedures (ESG and POSE) perform tissue folding to reduce stomach volume and therefore promoting weight loss [21, 22]. Another device in the first approach is the robotic suturing system - a master-slave flexible endoscopic system that has two flexible, 5-degree-of-freedom (DOF) robotic arms: a suturing instrument and a grasper [23, 24]. The high dexterity robotic arms enable surgeons to create running stitches and tie surgical knots endoscopically via a master console.

The second approach is the use of clips which are a type of simple, fast deployable device for endoscopic hemostasis and may be adapted for wound closure without using tendon-like surgical sutures. The size of the hemoclips is small enough to be delivered through endoscope channels. Their two jaws, when deploying, clamp the tissue to stop bleeding [25]. Another technology, over-the-scope clip (OTSC) (Ovesco Endoscopy AG, Germany) is a large nitinol clipping that provides high-force capture to compress the tissue securely for hemostasis and wall lesion closure [26].

The third approach is to improve conventional surgical sutures by incorporating advanced functional materials and new designs with enhanced mechanical properties. Although most surgical sutures for wound closure require secure knot tying with sufficient strength and desired compression to join separated tissues, the creation of a proper knot during the suturing process with sufficient tension is challenging. To avoid knot tying, conventional smooth sutures have been upgraded to barbed sutures that offer self-anchoring to tissues, eliminating the need for a knot [27]. Although barbed mechanisms have been introduced to effectively eliminate the need for complex knot tying and overcome the breakage problem, they still require a high level of suture pretension during the closure procedure before the anchors are deployed [28, 29]. A wide range of biomaterials have been used to produce advanced sutures including silk, linen fibers, nylon, polyester, polyurethane, or stainless steel with different surface textures (see [3, 4] for more details). Recent years have revealed an emerging trend towards functional sutures to provide the wound with some additional benefits [3]. Antimicrobial sutures are coated with
antimicrobial agents such as triclosan or silver nanoparticles to prevent postoperative infections [7, 30]. Furthermore, drug-eluting sutures can release drugs at a specific site to maximize the therapeutic effect [31]. Stem cell seeded sutures can provide biological scaffolds to speed up cell growth and thus accelerate tissue regeneration [32].

Another emerging tendency for wound closure is the use of programmable sutures. These sutures, a type of filament actuator or sensor, can monitor temperature, pH, or tension of the suture. Shape memory sutures that are made from shape memory polymers (SMPs) can induce self-tightening knots under thermal heated [2, 33, 34]. Although significant progress has been achieved, this technology requires thermal heating at the suturing site, making them an ill-suited candidate for use in practice. Smart electronic sutures such as the ones developed by Kim et al. are polymer strips with integrated flexible silicone temperature sensors and microheater to monitor wound temperature to support the healing process [6]. Although other electronic sutures with additional sensors to monitor pH, oxygen, temperatures, or wound exudates have been developed [3, 6, 35], all of them are not able to adjust tension such as knot tying and this has limited their use in practice.

There is a huge need for a new class of smart surgical sutures that can automatically tighten defect tissues with controlled tension while offering uniform force distribution within the tissue without the need for any human intervention. In this study, we introduce a proof-of-concept of a new smart surgical suture (S\textsuperscript{2} suture) made from a soft hydraulic artificial muscle. The core technology of the S\textsuperscript{2} sutures is the soft, scalable, and flexible tendon-like artificial muscle (STAM) which has been introduced in our previous work [36, 37]. Briefly, this soft hydraulic artificial muscle receives input fluid pressure to elongate to the desired length and accumulates potential elastic energy and subsequently generates contraction force, returning to its initial state when releasing the pressure. It can provide high elongation, high length-per-diameter ratio, miniaturization feasibility, and the ability to tune its initial stiffness or generated contraction force. The abundant availability of biocompatible microtubule and micro-coil materials accompanied by the simple fabrication method enables various applications of the STAM for smart sutures.

The S\textsuperscript{2} suture can adjust its length to achieve the desired tension at the time it is fabricated and automatically tighten its knot or deploy anchors to stabilize the suture against the sewed tissues without using any external pulling force, which is normally required in conventional sutures. In addition, our S\textsuperscript{2} sutures are also equipped with different anchors that can effectively secure
the defect tissues without the need for using knots, offering flexible choices to meet different demands of wound closure. We also describe the development and manufacture of these new S$^2$ sutures, and examine their capability of automatic knot tying and deployable anchors using ex-vivo experiments on fresh porcine stomach and colon. The S$^2$ sutures are potential candidates for both interrupted and running (continuous) stitches, thus making them useful for a wide range of applications such as skin wound closure, plastic surgeries, weight loss surgeries, and perforation closure for internal organs (Fig. 4.1).

**Figure 4.1.** Smart surgical suture with two typical surgical stitches and their potential application areas. The pictures of strabismus surgery, tendon repair, bone surgery, minimally invasive surgery for internal organs, cosmetic and reconstructive surgery, wound closure, and cervical corrections were “Designed by Freepik”.
4.3. Results

4.3.1. Smart surgical sutures (S² sutures)

In this work, we introduce a new class of soft artificial muscle-driven S² sutures that not only offer effective anchoring functions of both knot tying and barbed anchors but also provide desired and uniform tensile distribution without the need for additional intervention. For ease of comparison, the S² suture with knotting function is named “S² suture-knot” while that with barbed anchors is named “S² suture-anchor.” Typically, the S² suture-knot consists of a soft tendon-like artificial muscle (STAM), a pressure locking mechanism (PLM), and a commercial surgical needle (Fig. 4.2A). The STAM is a flexible, soft artificial muscle made from a miniature soft silicone tube inserted into a micro-coil so that it can be elongated to store elastic energy upon hydraulic pressurization and exert contraction force when releasing the pressure. Regarding the S² suture-anchor composition, both ends of the STAM are equipped with locking anchors (Fig. 4.2B). These anchors can be automatically deployed to secure the tissue without the need for a surgical knot, which requires complex manipulation of the closure device. One end of the STAM is connected to a PLM to hold and release its pressure. The other end of the STAM is connected to a cone-shaped suture tip and a curved surgical needle to facilitate the tissue puncture.

We utilize the contraction motion of the STAM after releasing the hydraulic pressure to automatically tighten the suturing knot or deploy anchors when all stitches are made. To do so, the STAM is first pressurized and kept in its stretched state by a PLM. One of the distinguishing features of the STAM compared to conventional sutures is that it provides uniform tension distribution along the artificial muscle body regardless of its length and configuration.

Both the S² suture-knot and S² suture-anchor are equipped with a PLM to hold the inner pressure of the STAM at a predetermined threshold to maintain the desired elongation. After making all stitches, the PLM is cut to release the pressure to shorten the STAM. When producing a STAM, a flexible fluid transmission tube is used to connect the STAM body to a fluid source. While input pressure from the fluid source to the STAM is maintained, the fluid transmission tube is locked and becomes a PLM. We propose three different PLM designs including a soft tube PLM (sPLM, Fig. 4.2C) made from soft rubber tubing, a heat seal tube PLM (tPLM, Fig. 4.2D) made from flexible polyethylene terephthalate (PET) tubing, and a hard tube PLM (hPLM, Fig. 4.2E) made from polytetrafluoroethylene (PTFE) tubing. Three types of PLMs require different locking methods: a simple overhand knot for the sPLM, heat
Chapter 4. Smart surgical sutures

seal effect and reinforced thread for the tPLM, and a cylindrical plug for the hPLM. The sPLM is easier to cut but has a relatively lower pressure threshold and thus smaller suture tension compared to the tPLM and hPLM.

![Smart surgical sutures (S² sutures)](image)

**Figure 4.2.** Structure of the smart surgical sutures (S² sutures). (A) The S² suture-knot can be knotted as conventional surgical sutures. (B) The S² suture-anchor formed by combining the S² suture-knot with three different types of anchors. (C-E) Design of different pressure locking mechanisms (PLMs) and their prototypes.

The anchors are responsible for holding separated tissues in place so that surgical knots can be eliminated, releasing surgeons from this arduous task, especially in confined spaces during endoscopic surgeries. We introduce three different designs for the anchors (Fig. 4.2B): 3D print, lantern, and sawtooth. The 3D printed anchors are made from hard plastic materials by commercial 3D printers. It has a cone shape with four barbs to facilitate one-way tissue
Figure 4.3. Fabrication, working concept, and prototypes. (A) (i) $S^2$ suture-anchor with 3D printed anchors at the initial phase (no pressure). (ii) Completed $S^2$ suture-anchor with pressurized artificial muscle. (iii) Making stitches through tissue. (iv) Tightening the suture and tissue by releasing pressure. (B) Deployment of the lantern and sawtooth anchors after releasing the pressure. (C) $S^2$ suture-anchor prototypes made from STAM, three types of anchors, and commercial surgical needles (details in Table 4.1). (D) Wound closure procedure using dual surgical robotic arms with continuous stitches of the $S^2$ suture-knot. (E) An $S^2$ suture-knot prototype at high and low pressure.
puncture and suture locking in the opposite direction. The lantern and sawtooth anchors are flexible plastic hollow tubes with patterned cuts so that they can be deployed to hold the separated tissues once the fluid pressure is released. A tube with longitudinal cuts (spare at two ends) produces a lantern-like shape when lengthwise compressing its two ends. The triangle cuts (sawtooth) create a bending anchor upon deployment where the STAM is shortened after hydraulic depressurization.

Table 4.1. Specifications of \( S^2 \) suture components.

<table>
<thead>
<tr>
<th>Micro-coil</th>
<th>Microtubule</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>STAM OD1.49</strong></td>
<td></td>
</tr>
<tr>
<td>OD 1.49 mm, wire 0.17 mm,</td>
<td>OD 1.19 mm, ID 0.64 mm,</td>
</tr>
<tr>
<td>maximum strain 416%, stainless steel,</td>
<td>maximum strain 786%, silicone rubber, durometer 50A</td>
</tr>
<tr>
<td><strong>STAM OD0.8</strong></td>
<td></td>
</tr>
<tr>
<td>OD 0.8 mm, wire 0.065 mm,</td>
<td>OD 0.61 mm, ID 0.31 mm,</td>
</tr>
<tr>
<td>maximum strain 482%, stainless steel,</td>
<td>maximum strain 855%, silicone rubber, durometer 50A</td>
</tr>
<tr>
<td>Anchor</td>
<td></td>
</tr>
<tr>
<td>3D print OD 3.6 mm, L 5 mm, 4 barbs, polylactic acid (PLA+), 3D print</td>
<td>Lantern OD 2 mm, ID 1.6 mm, L 20 mm, polyolefin, 3 cuts 120° apart</td>
</tr>
<tr>
<td>Sawtooth</td>
<td></td>
</tr>
<tr>
<td>OD 2 mm, ID 1.6 mm, L 20 mm, polyolefin, cutting angle 80°</td>
<td></td>
</tr>
</tbody>
</table>

The fabrication process and working concept of the \( S^2 \) suture-anchor with interrupted stitch are shown in Fig. 4.3A and the Methods section. The formation of the lantern and sawtooth anchors to secure the tissue is illustrated in Fig. 4.3B. To validate the new concept, we fabricated three \( S^2 \) suture-anchors with the proposed anchor designs (Fig. 4.3C). Detailed specifications can be found in Table 4.1. Each \( S^2 \) suture-anchor consists of a 70-mm-long STAM OD1.49 mm connected to anchors at both ends. Figure 4.3D illustrates the stitching steps for the \( S^2 \) suture-knot via dual surgical robotic arms and the main procedures to form a secured knot. The suturing procedure of the \( S^2 \) suture-knot with interrupted or continuous stitches is similar to those of conventional surgical sutures where the suture body will form a secured knot to tighten the separated tissues. However, instead of constantly maintaining the suture tension throughout the procedure and tying a tight knot in the case of using conventional surgical sutures, the \( S^2 \)
suture-knot implementation allows loose stitches and a loose knot. The suture tension and a tight knot to close and secure the wound will be achieved after releasing the hydraulic pressure. Similar to the $S^2$ suture-anchors, we also fabricated two different prototypes for the $S^2$ suture-knots (OD1.49 × L100 mm and OD0.8 × L100 mm, details in Table 4.1) where a surgical needle is directly connected to one end and the other end is equipped with an hPLM. Figure 4.3E shows an $S^2$ suture-knot prototype at high and low input pressure.

### 4.3.2. Experimental characterization

**Evaluation of the STAM capability**

We built a testing platform to characterize the STAM including hysteresis profiles of input volume and pressure versus output elongation, pressure limit, and elongation-force relationship. Details can be found in the Methods section. We produced three STAM specimens with similar specifications (STAM OD1.49, Table 4.1) but with different lengths of 30, 45, and 60 mm for the experiments. In the elongation experiments, we supplied a 0.2 Hz sinusoidal signal with different amplitudes to the syringe plunger so that each specimen could reach a maximum pressure of 1 MPa. Results (Fig. 4.4C,D) revealed that output elongation is proportional to both input volume and pressure. Furthermore, different muscle lengths required different input volumes (around 0.05, 0.06, and 0.07 ml for L30, L45, and L60 mm STAM, respectively) to reach 1 MPa. However, this 1 MPa pressure caused similar elongation to three specimens, 54.9% for L30, 53.2% for L45, and 52.2% for L60 mm STAM (standard deviation 1.37%). Hysteresis profiles show a smaller gap between pressurizing and releasing phases in the volume-elongation graph than that of the pressure-elongation graph, depicting that the specimens were more closely responding to volume changes than pressure. Next, we gradually increased input pressure to achieve a higher elongation (over 100%). Results in Fig. 4.4E,F show similar hysteresis profiles for the pressurizing phase compared to Fig. 4.4C,D. All three specimens did not burst when reaching the pressure sensor limit (1.85 MPa) and could resume their initial lengths. At 1.85 MPa input pressure, three specimens achieved an elongation of 107±2.1%.

We have investigated the STAM elongation-force relationship by both analytical and experimental approaches (Fig. 4.4G,H). In the force tests, one end of the STAM was kept static while the other end was pulled by the motorized linear slider (accompanied by a load cell) at a velocity of 10 mm/s until reaching an elongation of 200%. Three specimens with different lengths (30, 45, and 60 mm of STAM OD1.49) were tested, with five trials for each specimen,
Figure 4.4. STAM and soft tube characteristics. (A) Experimental setup to measure STAM elongation. (B) Setup to establish the STAM elongation-force relationship. (C, D) Hysteresis profiles of input (volume and pressure) and output elongation of three different STAM lengths, when supplying pressure until 1 MPa. (E, F) Same as C and D but with input pressure reaching pressure sensor limit (1.85 MPa). (G) Uniform tension distribution of a small segment of the STAM. (H) Elongation-force relationship of three different STAM lengths accompanied by the analytical modeling result (RMSE: root mean square error). (I) Setup to measure burst pressure of soft tubes and a table shows specimens’ specifications. (J) Burst pressure of three soft tube sizes.
and mean values plotted. Experimental results show a proportional relation between elongation and force (Fig. 4.4H) where insignificant deviation can be found between three STAM lengths. They achieved 2.87±0.05 N at 200% elongation.

Force equilibrium of a small STAM section is illustrated in Fig. 4.4G and described in Eq. (1) whereby the external force $F_{out}$ and the force $F_p$ caused by hydraulic pressure stretch the section while the micro-coil spring force $F_c$ and the microtubule elastic force $F_t$ tend to return the section to the initial state.

$$F_{out} = F_t + F_c - F_p$$

(1)

In the context of S² suture implementation where the hydraulic pressure is drained ($F_p = 0$) to trigger the contraction motion at the end of the suturing procedure, the STAM contraction force will reach the maximum $F_{out,max} = F_t + F_c$. This equation can be represented in Eq. (2) with detailed explanations presented in reference [36].

$$F_{out,max} = \alpha E A_t \left(1 - \frac{1}{1 + x/l_i}\right) + k_c x$$

(2)

where $E$ and $A_t$ are Young’s modulus and cross-section area of the microtubule, respectively; $k_c$ is the spring constant of the micro-coil; $l_i$ is the STAM initial length; $x$ is the STAM displacement when elongating; and stretch ratio $\alpha$ is the ratio between the microtubule length and micro-coil length.

In our previous study, the analytical model could effectively capture experimental data until the deflexion point ($\varepsilon = 100\%$) but showed poor performance at higher elongation. This was because $E$ was assumed to be a constant while it was augmented at high strain. This study improves the analytical model given in Eq. (2) by incorporating a dynamic Young’s modulus based on immediate elongation, which is described in Eq. (3). The equation consists of two constituents including the constant value $E_0$ (Young’s modulus at 100% strain) when elongation $\varepsilon \leq 100\%$ and a simple exponential function with the exponent $\varepsilon - 1$ when $\varepsilon > 100\%$.

$$E(\varepsilon) = \begin{cases} E_0, & \varepsilon \leq 100\% \\ E_0 e^{\varepsilon - 1}, & \varepsilon > 100\% \end{cases}$$

(3)

The dashed line in Fig. 4.4H represents the analytical model given in Eq. (2) with the following parameters: $\alpha = 0.9$, $A_t = 0.8026$ mm$^2$, $l_i = 60$ mm, $k_c = 0.01$ N/mm, $E$ follows Eq. (3) with $E_0 = 1.648$ N/mm$^2$, $x = \varepsilon l_i$ where $\varepsilon = 0-200\%$. The analytical model closely followed experimental
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data the entire elongation range with the maximum root mean square error (RMSE) of 0.0967 N.

Burst pressure of the pressure locking mechanism (PLM)

We first examined the burst pressure of the sPLM (soft tube PLM) using a similar setup in Fig. 4.4A. Detailed processes can be found in the Methods section. Three different sizes of soft rubber tubing (ST1, ST2 and ST3) were tested (Fig. 4.4I). Mean values and standard deviation of five trials for each tube size are presented in Fig. 4.4J. The sPLM ST1 and ST2 with a corresponding wall thickness of 0.45 and 0.4 mm could hold a pressure of 1.62 and 1.43 MPa, respectively, before bursting. The burst pressure of the ST3 (wall thickness 0.5 mm) exceeded the pressure sensor limit (1.85 MPa). With the S\textsuperscript{2} suture prototypes in Fig. 4.3C, if the required elongation is less than 60%, all three sizes of the soft tube can be used as an sPLM. We also tested the burst pressure of the tPLM and hPLM. The results revealed that they could surpass the pressure sensor limit before bursting. Therefore, the PLM and STAM were compatible with each other in terms of input pressure until the threshold of 1.85 MPa.

Anchors’ puncturing and holding force of the S\textsuperscript{2} suture-anchors

We also characterized the puncturing and holding force of the S\textsuperscript{2} suture-anchors. We created three different designs for the anchor specimens: sawtooth design, lantern design, and 3D printing design where their specifications are shown in Table 4.1. We also fabricated an S\textsuperscript{2} suture without anchors (a single STAM only) as a baseline comparison. The experimental diagram and actual setup can be found in Fig. 4.5A,B where a motorized linear slider, accompanied by a load cell, pulls the anchor specimens through a real porcine stomach tissue (~3 mm thickness, Coles supermarket, Sydney, Australia). The tip of each specimen is equipped with a miniature cone-shaped 3D printed block (see the S\textsuperscript{2} suture-anchor design) to ease the puncturing. The same procedure was applied for holding force characterization. However, the anchor specimens have been deployed and switched to the opposite direction.

In contrast, the S\textsuperscript{2} suture without anchor could provide both puncturing and holding force in one pulling motion. Mean values and standard deviation of five trials for each specimen (Fig. 4.5C) revealed that the S\textsuperscript{2} sutures with anchors have a larger puncturing force and a much higher holding force compared to the baseline (S\textsuperscript{2} suture without anchor). While the specimen without anchor required around 4.7 N to penetrate the porcine stomach, the sawtooth, lantern and 3D printed anchors needed around 7 N (1.49 times larger), 6.3 N (1.34 times larger), and 10.2 N (2.17 times larger), respectively. Regarding the holding force, the specimen without
anchor could hold only around 1.6 N whereas the sawtooth and lantern anchors could achieve holding forces of 7.6 N (4.75 times larger) and 9.2 N (5.75 times larger), respectively. Impressively, the 3D printed anchor specimen could hold 22.9 N (14.3 times larger) before perforating the tissue.

Figure 4.5. Anchor capability. (A) Experimental diagram to measure the puncturing and holding force of different types of anchors through the real porcine stomach tissue (~3 mm thickness). (B) Actual experimental setup. (C) Experimental results.

**Knot security and self-tightening capability of the S² suture-knot**

For wound closure during surgical procedures, tying knots with sutures is an essential component of maintaining tissue opposition where the security of a knot is crucial to hold separated tissues for promoting tissue healing. In this section, we will validate our hypothesis that the developed S² suture-knot could be applied loosely in its temporary shape during the suturing process. Once the temporary knot is created, the hydraulic pressure will be reduced so
that the suture will be shrunk to automatically tighten the knot with a pre-determined tension. We performed a set of experiments to test the feasibility of these concepts for our $S^2$ suture-knots including the ability to automatically tighten a knot with different loops and the knot security such as knots untying, or suture slipping from the clamps.

Figure 4.6. Self-tightening capability and knot security of the $S^2$ suture-knot. (A) A prototype (OD1.49 $\times$ L70 mm) is pressurized to 100% elongation and tied a loose knot with both ends are fixed. The knot is tightened when reducing input pressure. (B) Similar to A but with a prototype OD0.8 $\times$ L100 mm and both ends are set free. (C) Stability of the tightened knots after 1 week. These experiments are available in the Supporting Video.

As an illustration, we fabricated two different prototypes of the $S^2$ suture-knots (STAM OD1.49 and STAM OD0.8, Table 4.1). We elongated the $S^2$ suture-knot (OD1.49 $\times$ L70 mm) to about 100% elongation, formed a loose knot (Fig. 4.6A) and fixed both ends. We then withdrew the fluid (water) by slowly reducing the pressure via a miniature syringe. In this experiment, we did not perform the tension measurement of the $S^2$ suture-knot, but we instead examined its
self-tightening capability and knot security. Figure 4.6A shows how the suture knot is formed and automatically tightened when the STAM is shortened. We also conducted other experiments to examine the self-tightening capability of the knot and its security in the case when both ends of the suture were set free. Figure 4.6B shows the process of forming a tightened knot of an $S^2$ suture-knot (OD0.8 $\times$ L100 mm) once the fluid pressure is reduced. The results from Fig. 4.6 show that our $S^2$ sutures were able to self-shrink and tighten their knots with the desired tension (corresponding to pre-determined elongation, Fig. 4.4H). The secured knot was well maintained without occurring any failures such as slipping open or being untied after 1 week (Fig. 4.6C). These results also confirmed our hypotheses that the $S^2$ sutures could automatically tighten their knot without using any external human intervention.

4.3.3. Ex-vivo experimental validation

A major challenge for most wound closure procedures is the maintenance of high pulling force and uniform tension for the suture during the surgical procedure. This section will demonstrate proof-of-concept for the use of our $S^2$ sutures with uniform tensile distribution without the requirement of high pulling force in several surgical procedures. They include full-thickness defect closure or perforation closure, gastric plication for weight loss surgery, and proof-of-concept of cervical correction of the cervix. All procedures were performed on a fresh porcine colon and stomach (Coles Supermarkets, Sydney, Australia).

Perforation closure with the $S^2$ suture-anchors

We fabricated three prototypes of the $S^2$ suture-anchors using the same size STAM (OD1.49 $\times$ L70 mm). Each STAM was hydraulically pressurized and locked at 50% elongation and equipped with different anchors and surgical needles (Fig. 4.3C). We first created a straight 30-mm-long perforation on the fresh porcine stomach (Fig. 4.7A). Next, we manually made six running stitches by the $S^2$ suture around the perforation to close the wound using two pairs of surgical tweezers. Each stitch was made by puncturing the tissue with a surgical needle followed by the STAM body. We then cut the pressure locking mechanism to release the inner pressure which automatically deployed the anchors and then tighten the separated tissues. The entire perforation closure procedures for the $S^2$ suture-anchors with sawtooth are demonstrated in Fig. 4.7A and the Supporting Video. We also repeated the same procedures for the $S^2$ suture-anchors with lantern and 3D printed anchors. The results are shown in Fig. 4.7B,C. The successful closures for the perforation indicate that all three $S^2$ suture-anchor prototypes were
working as desired and could successfully achieve the required perforation closure requirement.

![Perforation closure and tissue folding procedure with the S\textsuperscript{2} suture-anchors.](image)

**Figure 4.7.** Perforation closure and tissue folding procedure with the S\textsuperscript{2} suture-anchors. (A) Perforation closure procedure with six running stitches by sawtooth anchor suture. (B, C) The same procedure applies to the lantern and 3D printed anchor sutures, respectively. (D) Tissue folding procedure (weight loss surgery) by 3D printed anchor suture. (E, F) The same procedure applies to the lantern and sawtooth anchor sutures, respectively.
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Tissue folding with the $S^2$ suture-anchors

Gastric plication is a minimally invasive weight-loss surgical technique that reduces the size of the stomach to promote weight loss [21, 22]. This technique does not have any implanted devices such as gastric banding or involve any removal of portions of the stomach. The tissue folding technique or gastric plication is a weight-loss surgical technique that normally requires the use of surgical sutures where high tension is required to tie a knot and to maintain a sufficient level of force to fold the tissue. In this technique, each fold is completed by making and tightening a bilateral stitch. To demonstrate the capability of our $S^2$ suture, we produced another three prototypes of the $S^2$ suture-anchors (similar to those in Fig. 4.3C but with shorter initial lengths of 30 mm). We designed the suture in a way that we can achieve a pair of stitches with a distance of around 36 mm that can fully fold the porcine stomach tissue (Fig. 4.7D). After double-puncturing (in and out) the tissue by the surgical needle, we pulled one anchor of the $S^2$ suture through the double-wall thickness tissue. We then cut the pressure locking mechanism to shorten the STAM length which subsequently formed a stomach fold. The surgical needle was then cut and removed. Figure 4.7D shows the entire tissue folding procedure using the 3D printed anchor suture. The tissue folding results of the lantern and sawtooth anchor are presented in Fig. 4.7E,F.

Perforation closure with the $S^2$ suture-knots

We also created two different prototypes of the $S^2$ suture-knots where their STAM had a diameter of 1.49 mm and 0.8 mm, respectively. Details of the characterization process can be found in the Methods section. Figure 4.8 shows the perforation closure results for the two prototypes of $S^2$ suture-knots on a fresh porcine colon. During the experiments, we observed that the $S^2$ suture-knots with the smaller diameter of STAM formed a secure and self-tightening knot more quickly than that of the larger STAM once the fluid pressure was reduced. This can be explained by the effect of the frictional surface of the outer micro-coil, its stiffness, and flexibility. In clinical practice, the lamination of soft biocompatible silicone or other polymers or coatings on the outer surface of the suture body is strongly recommended. This will contribute to a reduction in the surface friction of the STAM. In addition, a higher elongation of the STAM could promote the security of the knot tying.
Figure 4.8. Perforation closure with the $S^2$ suture-knots on a fresh porcine colon. (A) Results for the $S^2$ suture-knot OD1.49 × L70 mm. (B) results for the $S^2$ suture-knot OD0.8 × L100 mm. These demonstrations are available in the Supporting Video.

$S^2$ suture-knots for cerclage correction of the cervix

One of the main reasons for early opening of the cervix during childbearing is cervical insufficiency, meaning that the cervix is weak and unable to remain closed until the date of birth delivery [38, 39]. The correction of the cervix mainly involves the use of sutures to wrap around the cervix during pregnancy. Despite advances in the progress of medical technologies in the last few decades, the use of surgical sutures to temporarily treat recurrent pregnancy loss or premature birth still remains a major challenge in surgical procedures for cerclage correction. We demonstrate here that our $S^2$ suture-knot could be used as a potential candidate
to support the cerclage correction for the cervix without requiring any external pulling force while providing uniform tension to close the cervix. As a proof of concept, we fabricated a prototype of the S² suture-knot (OD0.8 × L100 mm). The S² suture-knot was pressurized to 60% elongation before wrapping around a soft cylindrical foam simulated as the human cervix. Once the S² suture-knot was fully wrapped around the soft foam and a loose knot was successfully created, we then quickly released the hydraulic pressure inside the STAM to self-tighten the knot. Figure 4.9 shows the experimental results for the proof-of-concept cerclage correction for the cervix. We visually observed a steady radial deformation of the soft foam under the S² suture-knot compression force. In addition, it could form a self-tightening knot once the fluid pressure was withdrawn.

Figure 4.9. Proof of concept for the cerclage correction of the cervix with the S² suture-knot. The prototype is pressurized to 60% elongation, wrapped around a soft foam, tied a knot, and finally released the fluid pressure to self-tighten and secure the knot.

4.4. Discussion and conclusion

Surgical sutures with self-tightening capability and uniform tensile distribution are highly desired in many surgical applications. The right tension in surgical sutures is an important factor to promote optimal tissue healing. A strong tension can restrict the blood flow and therefore create necrosis while a weak tension will result in the opening of incisions and thus the wound not healing properly. Tying a knot or providing high tension for sutures either in
interrupted or continuous stitches to close an open wound or to fold a tissue area during endoscopic surgery has long been a critical challenge for surgeons and suturing devices. Despite significant advances in robotic technologies for perforation closure, it is also critical and particularly complicated to precisely control the suture tension and automatically tighten the knot or anchor to close the wound lips with the right stress. This has been a major concern for surgeons for decades [4].

We have demonstrated proof-of-concept of new S² sutures that upon triggering can automatically tighten a loose knot or deploy anchors to close separated tissues. Although soft robotic technologies have emerged recently [40-48], to the best of our knowledge, there are no miniature soft fluid-driven actuators (diameter less than 0.8 mm, length longer than 1000 mm) that can be used as a surgical suture to perform the function of self-tightening a knot and anchoring as we have presented. One of the key components of our S² suture is a STAM that can elongate back and forth to induce uniform contraction force along its body under applied input hydraulic (water) pressure. A single STAM (OD1.49 mm) could hold a pressure up to 1.85 MPa without rupture and generate 107% elongation. At 200% elongation, it could produce a contraction force of 2.87 N. The STAM tension force is a combination of the micro-coil linear spring force and the microtubule nonlinear elastic force. These two forces are influenced by material properties and strain. In the case of uniformity of the micro-coil and microtubule materials, every small section like those in Fig. 4.4G can represent the force distribution of the whole STAM. When releasing input pressure, every STAM section experiences similar locally and simultaneously contraction motion to discharge elastic energy, leading to uniform tensile distribution of the whole S² suture.

We also introduced different pressure locking mechanisms (PLMs) including sPLM, tPLM, and hPLM to maintain the elongated state of the S² sutures during the suturing procedure and could be cut afterward to tighten the separated tissues. To secure the tissue, we also created three different types of anchors which relieve the surgeons from tying a complex surgical knot. Three designs for the anchors (3D printing, lantern shape, and sawtooth shape) have been designed to minimize puncturing force in one way and obstruct in the opposite direction which significantly raises the holding force. The sawtooth, lantern, and 3D printed anchor of the S² suture-anchors required 7 N, 6.3 N, and 10.2 N, respectively, to penetrate a 3 mm thick porcine stomach tissue. Their corresponding holding forces were 7.6 N, 9.2 N, and 22.9 N. It is noted that all three anchor specimens had larger outer diameters compared with a single STAM so that they required a higher force to puncture the tissue. The higher puncturing force compared
to holding force from the STAM with no anchors can be solved by using a smooth connector or a larger surgical needle. The sawtooth and lantern anchors required an equivalent puncturing force as they were made from the same flexible tube. However, after being deployed, the lantern-shaped anchor provided a larger holding force than the bending-shape of the sawtooth anchor. The rigid 3D printed anchor had the largest outer diameter causing the largest puncturing force amongst the testing specimens. On the other hand, the use of rigid materials with large diameter and barbs gave the 3D printed anchor a significant holding force. A suture tip and a commercial surgical needle were employed as a guide to facilitate the puncturing step.

All three $\text{S}^2$ suture-anchors have been successfully demonstrated functionality-wise in ex-vivo experiments where they could close a 30-mm-long perforation of a fresh porcine stomach by six running stitches. Also, the $\text{S}^2$ suture-anchors could create a tissue fold by tightening a bilateral stitch. Multiple tissue folds may be required in order to achieve the desired stomach volume reduction for weight loss control applications. It is noted that the perforation closure and tissue folding procedures were performed under an open surgery-like setting with no space restrictions and without dynamic movements of the tissue. Although we have not quantitatively evaluated the possible effects of high burst pressure on tissues yet, we hypothesize that it is unlikely that the burst pressure from our smart sutures will cause any harm to surrounding tissues. In our approach, the sutures carry a very small and fixed amount of fluid volume (<0.1 ml). When cutting the pressure locking mechanism, the inner pressure of the sutures drops rapidly and immediately, causing an instantaneous fluid jet to firstly hit the cutting instrument and dissipate smoothly and gently to the surrounding environment. In ex-vivo experiments, we also observed that there was no fluid jet coming to contact directly with the porcine tissues and therefore the harmfulness to the surrounding tissues are unlikely. However, we would recommend that more experiments to study the effect of high burst pressure on surrounding tissues should be carried out in future works before implementing the smart sutures in preclinical and clinical trials.

We also developed the $\text{S}^2$ suture-knot that is an $\text{S}^2$ suture without anchors. The $\text{S}^2$ suture-knots can be used as conventional surgical sutures. However, they could be automatically tightened, securing their knot without using any external human intervention or pulling force. In addition, the tied knot was able to hold its secured shape after one week without occurring any failures caused by slip or opening. To further validate the usefulness of the $\text{S}^2$ suture-knots for wound closure applications where knot tying is highly desired, we performed ex-vivo perforation closure experiments on the fresh porcine colon. We also conducted a proof-of-concept
experiment to demonstrate the use of our $S^2$ suture-knot for a complex surgical procedure – the cerclage correction of the cervix. All results confirmed that our new $S^2$ suture-knots were able to automatically tighten and secure the target tissues at desired suture tension, which is challenging for existing suture technologies.

It is noted that the current $S^2$ suture size is around 0.8 mm in diameter, which is still considered large compared to commercial sutures. In future work, smaller sizes of the suture (<0.5 mm in diameter) will be fabricated. In addition, all prototypes in this paper use a stretch ratio $\alpha = 0.9$-1 which is not an optimal design. We previously demonstrated that a reduction of the stretch ratio will increase its stored elastic energy and thus exert a higher contraction force. For the next phase of the project, we will also re-design the current STAM with relevant finite element analyses (FEAs) via COMSOL Multiphysics (COMSOL Inc., USA) in order to determine the desired parameters towards the achievement of the optimal smart suture. For further development, the $S^2$ suture-anchors can be also improved to some extent. Firstly, the $S^2$ suture-anchors have a relatively large size and larger puncturing force compared with conventional sutures, restricting their use in practice. Therefore, miniaturization of the anchor sizes is crucial for it to be used in clinical suturing procedures. Secondly, the current $S^2$ suture-anchors have a predefined distance between two anchors. As a result, the surgeons need to evaluate the perforation before choosing a certain length of $S^2$ suture. Although the developed $S^2$ sutures could not precisely control a precise value of the final tension, however they are able to provide a uniform tension distribution to close the wound that is impossible for existing sutures. As discussed in the introduction section, the uniform tension distribution for surgical sutures will facilitate the need of applying high tension for wound closure which is challenging in existing closure procedures including endoscopic surgery. In this paper, the $S^2$ sutures were designed in a way that they can provide a tension force within a desired range which is not too loose or too tight. Before suturing procedure, the $S^2$ sutures were pressurized to a certain initial elongation which corresponds to final tension once the pressure is released and the surgeons will perform closure procedure with stitching where a high pre-tension (which is highly desired in current surgical sutures) for the smart suture is not required before securing a know. It is noted the precise final tension of the $S^2$ sutures heavily relied on surgeons’ stitching skills. Therefore, future work should focus on a mechanism to adjust the anchor distance dynamically on-site to actively control the suture tension.

Both types of the $S^2$ sutures were made from non-absorbable materials such as stainless steel for the outer coil and silicone elastomer for the inner tube, a coating layer made from a
biocompatible material such as hydrogel or silk biomaterials which can carry drug or bacteria resistant elements around the suture body is highly desired. An investigation on other biocompatible materials that can be directly integrated into the $S^2$ suture components is essential before embarking on pre-clinical and clinical trials. In addition, the outer micro-coil will be upgraded by using a commercial suture wire as the main material. Finally, we also plan to refine, optimize, and combine the developed $S^2$ sutures with our recently developed flexible surgical endoscopic robots for in-vivo experiments on a living animal. Our developed $S^2$ sutures will also be a potential candidate for other surgical applications such as abdominal wound closure, hernia repair, sternal closure, or orthopedic procedures [49-52].

In conclusion, we have introduced the design and fabrication of a new class of smart surgical sutures that can be automatically tightened and secure the tissue after making stitches. Ex-vivo experiments have been successfully conducted to demonstrate the new suture capabilities for perforation closure and tissue folding. We believe that this proof-of-concept study will inspire more related improvement and development and benefit the medical research community.

4.5. Methods

4.5.1. Material selection and fabrication process of the smart sutures

The micro-coils used for all smart sutures are stainless-steel extension spring coils (Asahi Intecc Co., Ltd., Japan). The microtubules are silicone tubing (Saint-Gobain S.A., France). Firstly, a microtubule is inserted into a micro-coil to form a STAM with an initial length $l_i$. One end of the STAM is blocked while the other end is connected to a fluid source via a soft fluid transmission tube. Anchors are attached to both ends of the STAM. Secondly, the STAM is pressurized to the desired elongation $\varepsilon$ with corresponding new length $l_p$ ($l_p > l_i$). While input pressure from the fluid source to the STAM is maintained, the soft fluid transmission tube is tied using an overhand knot to lock the hydraulic pressure inside the STAM. The other end of STAM is connected to a suture tip and a surgical needle to form the $S^2$ suture-anchor. Thirdly, the $S^2$ suture-anchor punctures through the tissue to form surgical stitches. Fourthly, the pressure locking mechanism is cut by a surgical knife or scissors to release the hydraulic pressure which will shorten the STAM and automatically deploy the anchors to tighten the tissue. The same procedure is also applied to the other types of anchors (lantern and sawtooth). It is noted that two ends of each sawtooth anchor are required to adhere to the micro-coil when the STAM is in the elongated state in order to achieve the bending motion afterward. The 3D
printed anchors are made from polylactic acid (PLA+, Shenzhen Esun Industrial Co., Ltd., China) via a 3D printer (Ultimaker S3, Ultimaker B.V., Netherlands). The lantern and sawtooth anchors are manually fabricated from polyolefin tubes (heat shrink medical tubing, Nordson Medical, USA). The pressure locking mechanism (type sPLM) is a soft rubber tubing (McMaster-Carr Supply Co., USA). The suture tip is a 3D printed cone made from PLA+. The surgical needle is a nylon monofilament curved needle (Surgical Specialties Corporation, USA). All three \( S^2 \) suture-anchor prototypes were pressurized to 50% of elongation (with the corresponding length \( l_p = 105 \text{ mm} \)).

4.5.2. Testing setup for the smart suture characterization

The testing platform consisted of a motorized linear slider (Zaber, Canada) to provide input volume (distilled water) and pressure from a fluid source (medical syringe, BD Biosciences, Canada) to a STAM specimen (Fig. 4.4A). A pressure sensor (Honeywell, USA) was located right after the syringe outlet to measure input pressure. The specimen distal end was connected to an encoder (US Digital, USA) to measure output displacement. Figure 4.4B shows the experimental setup to establish the STAM elongation-force relationship. A load cell (Futek, USA) was used to continuously record the tension force. For the burst pressure characterization, the sPLM (soft tube PLM) was similar to that of Fig. 4.4A. We supplied hydraulic fluid (distilled water) to a soft tube, tied a knot and then connected it to the syringe outlet. Subsequently, the motorized linear slider pumped pressure to the soft tube until it burst, while the pressure sensor was continuously logging real-time pressure data.

4.5.3. Perforation closure with the \( S^2 \) suture-knots

Each prototype was directly connected to a surgical needle at one end of the STAM and the other end was connected to a hydraulic syringe via a miniature fluid transmission tube. The smart sutures were then elongated to 100% elongation (for the STAM with OD1.49 mm) and 60% elongation (for STAM with OD0.8 mm). A long perforation on a fresh porcine colon was created, then continuous stitches were performed around the wound, creating a loose knot (Fig. 4.8A,B). The fluid pressure was then released to shorten the STAM, which then induced self-tightening of the knot. Unlike experiments for the \( S^2 \) suture-anchors (Fig. 4.7) where the pressure locking mechanism was cut to rapidly reduce the hydraulic pressure, the hydraulic pressure was controlled via a miniature syringe in order to observe the dynamic motion of knot tying in real-time.
References


Chapter 5

Twisting and braiding artificial muscles

The soft hydraulic filament artificial muscle and its applications as a monofilament for a soft robotic glove and smart surgical sutures have been presented in Chapters 3 and 4. Inspired by human muscle consisting of multiple muscle fibres, this chapter investigates performance enhancement in terms of output elongation and force when twisting and braiding multiple artificial muscles. Several prototypes with different numbers and configurations of artificial muscles such as straight, twisted and braided are created and their performance is experimentally measured. Two analytical models are proposed to describe the relationship between elongation and force of twisted and braided artificial muscles. The implementation of these twisted and braided variants as artificial muscles to mimic human finger and elbow movements on 3D-printed exoskeleton models is demonstrated. The twisting and braiding abilities enable new possibilities. A soft tubular muscle is created using the hollow round braid technique. Its use is demonstrated in a number of applications including a compression sleeve for massage therapy, a hollow support device for medical applications and a tubular gripper for retrieving objects in confined spaces.

This work has been published:

5.1. Abstract

Research on soft artificial muscles (SAMs) is rapidly growing, both in developing new actuation ideas and improving existing structures with multifunctionality. The human body has more than 600 muscles that drive organs and joints to achieve desired functions. Inspired by the human muscles, this article presents a new type of soft artificial muscle fiber formed from twisting and braiding soft hydraulic filament artificial muscles with high aspect ratio, high strain, and high energy efficiency. We systematically investigated the relationship between input pressure and output elongation as well as contraction force of the new muscles using different configurations in terms of an array of single and multiple muscles arranged in nontwisting (or straight), twisting, and braiding variants. Experimental results revealed that the twisting and braiding configurations greatly enhanced the muscle elongation and generated force compared to their nontwisting/braiding counterparts. To demonstrate the new muscles’ usability, we implemented several muscle variants to bidirectionally manipulate 3D-printed human fingers and elbow, mimicking the human upper limb with a full range of motion. We also created a bio-inspired growing soft tubular muscle that could simultaneously exert longitudinal and radial expansion upon pressurization, similar to that of auxetic metamaterial structures. The new growing soft tubular muscles were experimentally validated and the results showed that they could be potentially implemented in several emerging applications, including smart compression garments, stent-like supporting devices, and tubular grippers for medical use.

Keywords: soft robotics, artificial muscles, muscle fibers, twisting and braiding, tubular muscle, tubular gripper, wearable devices

5.2. Introduction

The human body contains more than 600 muscles, which are used to actuate organs, joints, and tissues, to facilitate movement, move food through the digestive tract or allow the heart to pump blood to the lung and body [1]. Many studies have long been carried on to expand the use of artificial actuators that mimic biological muscles by directly using power (pressure, electrical, chemical, or thermal sources) to generate motion and force.

Soft artificial muscles (SAMs) have become an emerging research field in recent years thanks to their promising potential to revolutionize the traditional rigid robotic counterparts [2, 3].
With high versatility, flexibility, compliance, and resilience to disturbances [4, 5], the SAMs are ideal substitutes for conventional rigid actuators to develop human-friendly and safe robots [6], haptic devices [7], wearable exoskeletons [8, 9], and conformable robotic structures [10]. Abundant ideas in terms of actuation methods and conceptual designs of the SAMs have been proposed, including (i) electrically-driven muscles such as dielectric elastomer actuators [11, 12], electroactive polymers [13], and phase-change materials [14]; (ii) thermally-driven muscles such as shape memory polymers [15], shape memory alloys [16] and polymer fibers [17, 18]; and (iii) magnetically-driven muscles [19-21] and (iv) pressure-driven muscles [22-24].

Pressure-driven actuators offer great simplicity and performance in terms of high force and ease of implementation compared with others, making them the most popular used SAMs. Among them, pneumatic artificial muscles (PAMs) can be either operated in contractible or extensional mode, which is controlled by applying air pressure to a pneumatic bladder. The PAMs are lightweight because their main element is a thin membrane which allows them to be directly connected to the structure they power, an advantage to replace a defective muscle if they work in a group [2, 3, 22]. In an approximation of the human muscles, the PAMs are usually grouped in pairs, including one agonist and one antagonist.

The McKibben muscle, a type of PAM, was first developed in the 1950s for artificial limbs [8, 25]. Although this muscle offer many advantages such as high contraction force and low cost, the retraction strength of the McKibben is limited by the total strength of individual fibers in the woven shell, while the tightness of the weave limits the exertion distance. In an attempt to re-engineer the PAMs, other studies, which are not limited to the use of air pressure, have been introduced to optimize the performance of the PAMs. They include air-driven origami-inspired artificial muscle (FOAMs) [26], split tube pneumatic actuators [27], hydraulically fluidic fabric muscle sheets [28], vacuum-actuated muscle-inspired pneumatic structures [29], and inverse PAMs [30].

Although significant progress has been achieved, the insufficient contraction ratio poses a challenge for many fluid-driven muscles, including the PAMs and our recent new muscles, making them a disadvantage compared to the human muscles [25, 27]. To further minimize the McKibben muscles’ size, Koichi’s group developed miniature McKibben muscles where they inserted a miniature elastic bladder inside an outer braided sleeve. The new muscles could produce a high contraction elongation or force when receiving positive pneumatic pressure [25,
To increase the muscle elongation limit, they grouped multiple single muscles in a braiding configuration, which subsequently resulted in an increase of the muscle contraction limit (28–41%) [33]. In another approach, Kurumaya *et al.* developed a three-strand braided active textile muscle which could increase the muscle elongation up to 26.8%, compared with those of multiple single McKibben muscles arranged in a straight configuration [32]. These results reveal that the McKibben muscles, once braided, could enhance their elongation limits and therefore their generated force.

The idea of twisting artificial muscles to increase their strength also attracted considerable attention recently [34, 35]. Using thermal heating as a driving source, Kharal *et al.* created composite bijels (bicontinuous interfacially jammed emulsion gels) by hydrodynamic twisting multiple bijel fibers around a core polymeric support fiber [35]. The composite bijels could increase the tensile strength up to 20 MPa, roughly 4000 times higher than that of the liquid bijel fibers. Many researchers have also investigated hybrid yarn artificial muscles (HYAMs), in which they tailored the multiple fibers in order to increase the muscle energy densities. Lima *et al.* introduced carbon nanotube (CNT) HYAMs consisting of stimuli-responsive guest-filled paraffin waxes into twist-spun CNT yarns [36]. This combination enables the customization of muscle stimulants with high contraction stress (up to 84 MPa). In a similar approach, Mu *et al.* used the stimuli-responsive material where the guest-filled CNT yarns were replaced by a sheath on a twisted or coiled core using inexpensive yarns [37]. The sheath-run muscles could generate a dense contractile power of 1.98 W/g, 40 times larger than those of the human muscle.

Yuan *et al.* created high-energy microengines from twisted shape-memory nanocomposite fibers [38]. The fiber is a synthesis of polyvinyl alcohol and graphene oxide nanoparticles. This HYAM could produce a work capacity of around 2.8 kJ/kg upon thermal exposure. Kanik *et al.* fabricated bimorph fibers from high-density polyethylene and cyclic olefin copolymer elastomer by the fiber-drawing technique [39]. The twisted bimorph fibers could achieve a work capacity of 7.42 kJ/kg when thermally stimulated. These results demonstrate that the artificial muscles, once twisted or braided, could enhance their elongation and strength, which can be implemented in several robotic applications.

We recently introduced soft microtubule artificial muscles (SMAM) formed from inserting a soft microtubule into a hollow microcoil as constraint layer [40]. The insertion method avoids manual wrapping of inextensible fibers along the microtube and hence allows a uniform distribution of the constrained outer coil to prevent unexpected failure when operating against
high hydraulic pressure. Using this fabrication method, the soft muscle size could be scaled down to 0.8 mm in outer diameter and could reach a high aspect ratio (~5000 of length:diameter). In addition, it also offers a high strain (at least 250%) and tuneable generated force using a prestretch ratio for the inner silicone tube while maintaining high energy efficiency and wide bandwidth (at least 20 Hz) [7, 40].

In this article, we introduce a new method of twisting and braiding multiple SMAMs, which can contribute to enhance the muscle elongation and hence contraction force. To demonstrate the effectiveness of the new approach, we constructed different prototypes for new SMAMs-based soft muscles using different configurations, including multiple single muscles arranged in nontwisting (or straight), twisting, and braiding variants. We experimentally characterized and compared the new muscle-generated strain and force corresponding to input pressure for each configuration. To better describe their elongation/force relationship, we also developed analytical models for the twisting and braiding configurations. We also showed that by braiding multifilaments with different turns and structures, we could form a new type of SAM that has enhanced capabilities to perform the desired actuation.

To validate our new concepts, we fabricated different prototypes and deploy them to manipulate knuckle joints of an index finger model and control an elbow joint of an arm model, mimicking the function of human muscles. We also created a newly bioinspired soft tubular muscle and demonstrate its multifunctions as a compression sleeve for massage therapy, which can be used to improve blood circulation or a stent-like hollow muscle that can be utilized to support the arteries or gastrointestinal (GI) tract while performing endoscopic surgeries. Finally, we braided multiple SMAMs to form a soft tubular gripper that could perform grasping tasks or retrieve objects in a confined space such as the human GI tracts or hollow tubes where conventional grippers find difficult to perform.

5.3. Soft microtubule artificial muscles

As shown in Fig. 5.1, we use SMAMs, which have an inner silicone microtube or microtubule and an outer constrained microcoil, for our proposed approach. Briefly, the SMAM is created by inserting this microtube into the microcoil (or extension microspring). The distal end of the microtubule is tied in a knot before sealing to the microcoil by an adhesive silicone, while the proximal end is connected to a fluid source via a fluid transmission tube [7, 40]. The
microtubule is a soft and high strain hollow tube made from silicone elastomer or silicone rubber. The spring coil comes from various materials such as stainless steel, brass, or polymer wires. The microcoil at the outer layer constrains the radial expansion of the microtubule under fluid pressure, resulting in a longitudinal lengthening of the SMAM. The SMAM in a form of filament can be made excessively long, which can be wrapped around a spool pin and will be cut to the desired length afterward (Fig. 5.1). Multiple muscle filaments can be twisted and braided to form an integrated muscle for further application or customization. Detailed material selections, frequency response, and muscle size can be found in Phan et al. [40].

The SMAM working principle is mainly based on the fluid dynamics of force transmission and the energy conversion from fluid pressure to mechanical work. Both the microtubule and the microcoil are elastic elements that can be deformed to generate mechanical strain and then store elastic energy in the form of an elastic force, which is lately discharged to reform their initial structure (Fig. 5.2A). A typical working cycle of the SMAM consists of three distinct phases: (i) initial phase where the muscle is in its relaxed position with no input pressure and no stored elastic energy; (ii) pressurizing phase where the muscle receives fluid pressure to induce the muscle elongation and thus store the elastic energy. The more pressure the muscle receives, the more elongation it exhibits, resulting in the more elastic energy it accumulates; and (iii) releasing phase where the fluid pressure is withdrawn. The SMAM always tends to discharge the stored elastic energy to restore its initial length. Once the fluid pressure is reduced, the SMAM will release its stored elastic energy and then convert it into mechanical work if the SMAM is working against an external load. A higher stiffness of the outer helical coil and higher Young’s modulus of the inner elastic materials together with a smaller stretch ratio of the inner microtubule will result in a stronger contraction force of the SMAM.

The relationship between the displacement $x$ and contraction force $F_{out} = F_{SMAM}$ of a single SMAM has been given in Phan et al. [40], which is now reintroduced here:

$$F_{out} = F_{SMAM} = \alpha E A_0 \left(1 - \frac{1}{1 + x/l_i}\right) + k_c x$$

where $\alpha$, $E$, $A_0$ represents the stretch ratio, Young’s modulus, and cross-sectional area of the microtubule, respectively; $k_c$ is the stiffness coefficient of the microcoil and $l_i$ is the initial length of the SMAM.

When combining $n$ SMAMs ($n>1$) in a straight configuration (or S-SMAM), the S-SMAM elongation is equal to that of a single SMAM under an assumption that each individual SMAM
has a similar physical structure. The S-SMAM will then exert a contraction force of \( nF_{SMAM} \) where \( F_{SMAM} \) is the contraction force of an individual SMAM in the S-SMAM. This new generated force can be expressed by the following:

\[
F_{S-SMAM} = nF_{SMAM} = n \left[ \alpha E A_0 \left( 1 - \frac{1}{1 + x/l_i} \right) + k_c x \right]
\]  

(2)

**Figure 5.1.** A new class of SMAMs inspired by the human skeletal muscles consisting of multiple muscle fibers. The SMAM fabrication steps include the insertion of a soft microtubule into a spring coil to create a long premuscle filament that can be cut to the desired length afterward. Multiple SMAMs can be twisted or braided to form a new class of artificial muscles with the desired structure such as tubular shape. SMAMs, soft microtubule artificial muscles.
5.4. Enhanced performance of the SMAMs with twisting and braiding configurations

To enhance the SMAM elongation while retaining its flexibility, we introduce here twisting and braiding configuration methods to form a new class of SAMs. Although prior studies showed that the strain limit of the SAM could be increased by braiding [33], however, twisting multiple fluid-driven artificial muscles to enhance its elongation limits is not widely implemented as we propose here. We demonstrate that the change of twisting or braiding angle or the number of twisting or braiding turns will affect the muscle elongation and thus its contraction force. We also further demonstrate that the new configurations of the SMAMs will enable tremendous possibilities to customize the muscle structure based on the required strain, stress, compactness, and flexibilities, to meet specific circumstances, which can be found in the next sections.

5.4.1. Twisting soft microtubule artificial muscle

The twisting soft microtubule artificial muscle (T-SMAM) can be achieved by twisting multiple SMAMs or folding a single SMAM to form multifolded segments and then twisted them together. In particular, the T-SMAM can be created by a combination of the two methods. There are different types of T-SMAM configurations, depending on the number of segments after folding, the number of twisting turns, and the number of participated SMAMs. Although the working principle of the T-SMAM is similar to that of a single SMAM, the muscle elongation can be tuned by varying the number of twisting turns, which subsequently increases its contraction force. It means that the T-SMAM with a higher extension ratio will produce a higher contraction force for the same initial configuration and nominal length compared to those smaller ones. For the T-SMAM, when pressurizing, besides the intrinsic longitudinal elongation of each individual SMAM segment, there will be a reduction in twisting angle, leading to an additional prolongation of the muscle length.

The change in the output elongation under the applied fluid pressure of the T-SMAM is given in Fig. 5.2B. Assume that the new muscle (T-SMAM) consists of \( n_t \) similar SMAMs and each individual SMAM has the same initial length \( l_{it} \), outer diameter \( d_i \), and twists \( m_t \) turns in the same direction. The T-SMAM will then have an outer diameter \( D_{it} \) and an initial length \( h_{it} \). Each SMAM will have the same pitch \( p_{it} (p_{it} = h_{it}/m_t) \) and twisting angle \( \varphi_{it} \) that can be obtained from a one-revolution decomposition in Fig. 5.2B and expressed by the following:
Figure 5.2. Working principle and analytical model illustration. (A) Input pressure generates elongation and force in a single SMAM. (B, C) Geometric parameters of twisting and braiding configurations, respectively, at the initial and pressurizing phase, accompanied by a one-revolution decomposition.
\[
\varphi_{it} = \tan^{-1}\frac{\pi(D_{it} - d_t)}{p_{it}} = \tan^{-1}\frac{\pi(D_{it} - d_t)m_t}{h_{it}}
\]

Under a fluid pressure \( P \), each individual SMAM will elongate with an amount of \( x_t \) to reach a new length \( l_{pt} \). This will lengthen the T-SMAM to a value of \( h_{pt} \). In practical applications, both ends of the T-SMAM should be maintained in their original configuration, meaning that they do not have any relative rotary motions. This condition contributes to maintaining a constant number of twisted turns \( m_t \) while the T-SMAM is in operation. However, the twisting angle decreases if the muscle elongates, inducing a closer contact between individual muscles and thus reducing the T-SMAM diameter from \( D_{it} \) to \( D_{pt} \). This radial shrinkage is defined by \( \varepsilon_{rt} = 1 - D_{pol}D_{it} \) and can be obtained by experiments. The parameter \( \varepsilon_{rt} \) mainly depends on the number of participated muscles \( n_t \), the muscle length \( l_{it} \), and the number of revolutions \( m_t \). In the pressurizing phase where the T-SMAM is under fluid pressure, it has a new twisting angle \( \varphi_{pt} \), length \( h_{pt} \), and pitch \( p_{pt} (p_{pt} = h_{pol}m_t) \), which can be described by the following:

\[
\varphi_{pt} = \sin^{-1}\frac{\pi(D_{pt} - d_t)}{g_{pt}} = \sin^{-1}\frac{\pi[D_{it}(1 - \varepsilon_{rt}) - d_t]m_t}{l_{pt}}
\]

\[
h_{pt} = l_{pt}\cos\varphi_{pt}
\]

From these equations, one can obtain the longitudinal elongation or \( \varepsilon_t = h_{pt}/h_{it} - 1 \) of the T-SMAM. The accumulated displacement \( x_t \) of each muscle can be expressed as follows:

\[
x_t = l_{pt} - l_{it} = \frac{h_{pt}}{\cos\varphi_{pt}} - \frac{h_{it}}{\cos\varphi_{it}}
\]

While the exerted force of each individual SMAM can be expressed by Eq. (1), the contraction force \( F_{out,t} \) of the T-SMAM is a summation of the total exerted forces of all participating SMAMs, which is given by the following:

\[
F_{out,t} = n_t\left[\alpha E A_0 \left(1 - \frac{1}{1 + x_t/l_{it}}\right) + k_c x_t\right]\cos\varphi_{pt}
\]

5.4.2. Braiding soft microtubule artificial muscle

Several segments of a single SMAM or multiple SMAMs or multiple T-SMAMs can be braided to create a braiding soft microtubule artificial muscle (B-SMAM). The B-SMAM structure can readily vary from traditional constructs that include the classic three-strand braid, multiple-strand flat, and round braid [41]. The working principle of the B-SMAM is similar to that of a
Chapter 5. Twisting and braiding artificial muscles

single SMAM or the T-SHAM. However, the elongation of the B-SHAM can be tuned by varying the number of braiding turns. The B-SHAM will have a higher extension ratio or higher contraction force for the same initial configuration and nominal length if a higher number of braiding turns are used during the fabrication. Similar to the T-SHAM, the B-SHAM always has a reduction in the twisting angle, resulting in a higher elongation of the muscle compared to that of an individual SMAM.

Round braid is a basic braiding technique that incorporates both twisting and weaving of associated filaments to form a rod-like product [41]. This technique typically requires an even number of participating filaments, in which they are divided into two groups. The filaments of the two groups will then twist in opposite directions and interlace wherever a cross occurs. Here, we introduce an analytical model for the B-SHAM using the same technique as that of the T-SHAM. Figure 5.2C illustrates the working principle of a B-SHAM made from $n_b$ single SHAMMs where each has an initial length $l_{ib}$, an outer diameter $d_b$, and $m_b$ twisting turns while weaving with its peers. The B-SHAM at the initial phase has a length $h_{ib}$ and an outer diameter $D_{ib}$. By decomposing one turn of a single B-SHAM, we can reveal two distinct features of twisting and weaving configurations: (i) twisting properties are depicted by pitch $p_{ib}$ ($p_{ib} = h_{ib}/m_b$) and twisting angle $\varphi_{ib}$; and (ii) weaving effect is represented by the interlacing angle $\theta_{ib}$. These initial parameters can be obtained from a one-revolution decomposition in Fig. 5.2C:

$$\varphi_{ib} = \tan^{-1} \frac{\pi(D_{ib} - 2d_b)}{p_{ib}} = \tan^{-1} \frac{\pi(D_{ib} - 2d_b)m_b}{h_{ib}}$$

In the round braid technique with $n_b$ muscles, each muscle will interlace two times with $n_b/2$ muscles of another group in every revolution. Therefore, each individual muscle will have $n_b$ interlacing points at each turn. The interlacing angle can be calculated by:

$$\theta_{ib} = \tan^{-1} \frac{d_b}{g_{ib}/n_b} = \tan^{-1} \frac{n_b d_b m_b \cos \varphi_{ib}}{h_{ib}}$$

where, $g_{ib} = \frac{p_{ib}}{\cos \varphi_{ib}} = \frac{h_{ib}}{m_b \cos \varphi_{ib}}$

Under a fluid pressure $P$, each muscle will elongate an amount of $x_b$ to reach a new length $l_{pb}$ or the B-SHAM will lengthen to $h_{pb}$. Similar to the T-SHAM, we assume that there is no relative rotation between two ends of the B-SHAM or the twisting turns $m_b$ are constant during the B-SHAM operation. At the pressurizing phase with the pressure $P$, the B-SHAM has a
new set of parameters, including interlacing angle $\theta_{pb}$, twisting angle $\varphi_{pb}$, length $h_{pb}$, and pitch $p_{pb} = h_{pb}/m_{b}$, which can be expressed by the following:

$$\theta_{pb} = \sin^{-1} \frac{n_b d_b m_b}{l_{pb}}$$  

$$\varphi_{pb} = \sin^{-1} \frac{\pi [D_{pb} - 2d_b]}{g_{pb}} = \sin^{-1} \frac{\pi [D_{ib}(1 - \varepsilon_{rb}) - 2d_b]m_b}{l_{pb} \cos \varphi_{pb}}$$  

$$h_{pb} = l_{pb} \cos \varphi_{pb} \cos \theta_{pb}$$

where $\varepsilon_{rb} = 1 - D_{pb}/D_{ib}$ denotes the radial shrinkage of the B-SMAM (from $D_{ib}$ to $D_{pb}$).

The longitudinal elongation of the B-SMAM will be $\varepsilon_{lb} = h_{pb}/h_{ib} - 1$ while the accumulated displacement $x_{b}$ of each individual SMAM can be expressed by the following:

$$x_{b} = l_{pb} - l_{ib} = \frac{h_{pb}}{\cos \varphi_{pb} \cos \theta_{pb}} - \frac{h_{ib}}{\cos \varphi_{ib} \cos \theta_{ib}}$$

It is noted that the exerted force of each individual SMAM is described by Eq. (1). The contraction force $F_{out,b}$ of the B-SMAM is a summation of the exerted forces of all participating SMAMs, which can be detailed as follows:

$$F_{out,b} = n_b \left[ \alpha E A_0 \left( 1 - \frac{1}{1 + x_b/l_{ib}} \right) + k_c x_b \right] \cos \varphi_{pb} \cos \theta_{pb}$$

### 5.5. Experimental characterization

In this section, we introduce the detailed experimental setup, fabrication of different muscle prototypes, and characterization of the new muscles together with model validations.

#### 5.5.1. Experimental setup and prototypes

We built an experimental platform (Fig. 5.3) to characterize the muscle performance, in which we aim to establish the relationship between input (volume and pressure) and output (elongation and contraction force). We then validated our developed analytical models given by Eqs. (1) to (14). We also compared the performance of different muscle prototypes to provide a better understanding of how the muscle parameters affect their elongation and their generated force.
Figure 5.3. Experimental setup to characterize muscle performance. (A) Elongation measurement. (B) Force measurement.

Figure 5.4. Specimens involved in the experiments (details shown in Table 5.1). The orange circles in each sub-figures represent the arrangement of SMAMs at the proximal end and the distal end to form desired muscle configurations.
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The experimental platform consists of a motorized linear slider (Zaber, Canada), which can provide input volume (pressure) to the muscle via a miniature syringe (BD Biosciences, Canada) and fluid transmission microtubes. A pressure sensor (Honeywell, USA) is located right after the syringe to monitor the fluid pressure. While the muscle’s proximal end was fixed, its distal end was connected to an encoder (US Digital, USA) to monitor the muscle elongation (Fig. 5.3A) or to a load cell (Futek, USA) to collect force data (Fig. 5.3B). We used a linear slider (Misumi, USA) as a guide to preventing the relative rotary motion of the two muscle ends. An elastic string maintained tension between the muscle’s distal end and the encoder, preventing slack when acquiring the displacement signal. Distilled water was used in this instance, although it should be noted that our SMAMs were developed to work with any liquid such as hydraulic oils. The use of these oils or higher boiling point fluids such as cooking oil or hydraulic oil avoids the boiling problem due to cavitation or when the SMAMs are used in high-temperature environments.

Table 5.1. Specifications of specimens

<table>
<thead>
<tr>
<th>Specimens</th>
<th>n</th>
<th>Length (mm)</th>
<th>m</th>
<th>D_i (mm)</th>
<th>ε</th>
</tr>
</thead>
<tbody>
<tr>
<td>SM1</td>
<td>Single</td>
<td>1</td>
<td>75</td>
<td>0</td>
<td>N.A</td>
</tr>
<tr>
<td>SM2</td>
<td>Straight</td>
<td>4</td>
<td>75</td>
<td>0</td>
<td>N.A</td>
</tr>
<tr>
<td>SM3</td>
<td>Twisting</td>
<td>4</td>
<td>70</td>
<td>3</td>
<td>4</td>
</tr>
<tr>
<td>SM4</td>
<td>Braiding</td>
<td>4</td>
<td>70</td>
<td>3</td>
<td>6</td>
</tr>
<tr>
<td>SM5</td>
<td>Straight</td>
<td>7</td>
<td>75</td>
<td>0</td>
<td>N.A</td>
</tr>
<tr>
<td>SM6</td>
<td>Twisting</td>
<td>7</td>
<td>70</td>
<td>3</td>
<td>5</td>
</tr>
<tr>
<td>SM7</td>
<td>Braiding</td>
<td>7</td>
<td>70</td>
<td>2</td>
<td>7</td>
</tr>
</tbody>
</table>

Fundamental muscles, \( l_i = 75 \text{ mm}, \ \alpha = 0.9 \)

<table>
<thead>
<tr>
<th>Microtubule</th>
<th>Spring coil</th>
</tr>
</thead>
<tbody>
<tr>
<td>Silicone rubber OD = 1.19 mm, ID = 0.64 mm</td>
<td>Stainless steel OD = 1.49 mm, ID = 1.1 mm</td>
</tr>
<tr>
<td>( E(100%) = 1.648 \text{ MPa} )</td>
<td>( k_c = 0.013 \text{ N/mm (length 50 mm)} )</td>
</tr>
<tr>
<td>Durometer 53A</td>
<td></td>
</tr>
</tbody>
</table>
We produced seven specimens, including the ones with straight, twisting, and braiding configurations (SM1 to SM7, Fig. 5.4) for experimental characterization. All specimens were constructed from the same fundamental SMAMs made from silicone rubber microtubules (Saint-Gobain, France) and stainless steel spring coils (Asahi Intecc, Japan). Detailed specifications of specimens can be found in Table 5.1. Each end of the specimen was attached to its 3D-printed blocks, forming a modular sample for easy installation and disassembling on the testing platform. We also used similar one-to-multi fluid distributors (made by 3D printer: Ultimaker, Netherlands; material: polylactic acid, Cubic Technology, Australia) to dispense fluid from a single syringe outlet to multiple muscles simultaneously. This ensured a similar energy loss amongst the specimens.

5.5.2. Elongation characterization

To analyze the muscle specimens’ performance, we applied 0.2 Hz sinusoidal signals to the syringe plunger in the elongation tests, providing input volume and pressure to the specimens. Amplitudes of these input signals varied for each specimen so that they could reach maximum pressure at the vicinity of 1 MPa. Experimental results (Fig. 5.5A, B) show that a single 75-mm-long muscle (SM1) required around 0.06 mL fluid volume to reach 1.05 MPa and achieve 48% elongation. Proportionally, to achieve the same pressure, four-strand specimens (SM2, SM3, SM4) and seven-strand specimens (SM5, SM6, SM7) needed ~ 0.23 and 0.4 mL of the fluid, respectively. However, while the two straight configurations (four-strand straight and seven-strand straight) achieved the same elongation as those of the single specimen, their twisting and braiding configurations produced slightly higher elongations of 53% for twisting (10.4% enhancement) and 55% for braiding (14.6% enhancement).

These elongation enhancements agree with our hypothesis in the previous section that the twisting and braiding configurations provide additional displacement by reducing the twisting angle in the pressurizing phase. Hysteresis relationships between input (volume and pressure) and output elongation can be identified in the corresponding graphs (Fig. 5.5A, B), where a gap existed between the pressurizing and releasing phase of each specimen. The underlying mechanism of this phenomenon is energy loss. The muscle discharges a smaller amount of energy in the depressurizing phase than it receives in the pressurizing phase. Furthermore, the incomplete removal of air bubbles inside the transmission fluid tubes and the muscles at the fabrication stage also widened the hysteresis gap. As such, the higher volume that the specimen requires, the larger the gap that occurs (Fig. 5.5A).
Figure 5.5. Experimental results of seven specimens. (A, B) Output elongation relates to input volume and pressure, respectively. (C, D) Hysteresis profiles of output contraction force and input volume and pressure, respectively. (E) Relationship between elongation and contraction force compared with analytical models. (F) Unit force comparison, where force values equal the specimen contraction forces divided by the corresponding number of participatory muscles.
5.5.3. Force characterization

We also carried out experiments to evaluate the generated force of the specimens. We first supplied fluid volume to the specimens until they reached 0.9 MPa then connected their distal ends to a load cell for the force measurement. Subsequently, we withdrew the fluid volume using 0.2 Hz sinusoidal signals from the linear actuator to induce specimen-dependent amplitudes where the pressure was reduced to 0.1 MPa. Results (Fig. 5.5C, D) reveal that after receiving 0.9 MPa in the pressurizing phase, a single 75-mm-long muscle (SM1) could exert ~0.69 N contraction force when releasing the pressure. In similar conditions, the four-strand straight and seven-strand straight specimens achieved forces of 2.73 N and 4.81 N, respectively. The twisting and braiding configurations of both four-strand and seven-strand specimens, as expected, reached a higher contraction force of 3.06 N (12.1% enhancement) for four-strand twisting, 3.17 N (16.1% enhancement) for four-strand braiding, 5.35 N (11.2% enhancement) for seven-strand twisting, and 5.53 N (15% enhancement) for seven-strand braiding.

Force enhancement is an inevitable consequence of elongation enhancement. As a result, the twisting and braiding configurations will achieve higher elongation than straight configurations at the same input pressure. In other words, they accumulate more elastic energy, resulting in the exertion of a stronger contraction force.

Hysteresis profiles of the input volume and pressure versus output contraction force (Fig. 5.5C, D) show the inverse relationship between the SMAM contraction force and the input volume and pressure. The hysteresis profiles of the pressurizing and releasing phases of each specimen are almost overlapping, which means that these specimens have consistent durability. This behavior reflects the immediate response with minor energy loss of the SMAM when it receives prestored elastic energy at the initial condition. Specimens in force characterization were continuously stretching. In contrast, specimens interchanged states (from relaxing to stretching and vice versa) in the case of elongation tests. Another interesting feature of the SMAMs is the linear relationship between the input pressure and output force (shown in Fig. 5.5D). This feature will allow us to achieve the desired generated forces on the proviso that the pressure is precisely controlled.

5.5.4. Elongation – force relationship

Experimental results in the previous parts enable us to interpret the proportional relation between elongation and contraction force of the SMAM via input pressure. We conducted
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experiments to establish a direct relation between elongation and force and validate our developed analytical models given by Eqs. (1) to (14). Using the previous specimens and testing platform, we gradually increased input volume and pressure so that we could collect contraction force at each blocked elongation ranging from 0% to 100% with a 10% increment. The average force and standard deviation of five testing cycles are presented in Fig. 5.5E. The chart reinforces the previous interpretation regarding the proportional relationship between contraction force and elongation and provides insight – similar curved patterns for all seven specimens.

The more elongation the muscle accumulates in the pressurizing phase, the more contraction force it exerts in the releasing phase. The spline curves are combinations of the spring coil’s linear characteristic and the microtubule’s nonlinear attribute. The curves are slightly convex until the inflection point (about 70% elongation), followed by a rapid increase due to the augmentation of the microtubule Young’s modulus. Figure 5.5E also illustrates analytical models of single SMAM [Eq. (1)], S-SMAM [Eq. (2)], T-SMAM [Eq. (7)], and B-SMAM [Eq. (14)], where parameters are given in Table 5.1. Our proposed models closely captured experimental data for all specimens until the inflection point (root mean square error ranging from 0.032 for SM1 to 0.3048 for SM6).

We divided each specimen’s experimental force data (Fig. 5.5E) by its corresponding number of individual muscles to obtain the unit force presented in Fig. 5.5F. There are two distinguishable groups, where the group with single and straight configurations has a larger unit force than those with twisting and braiding variants. The smaller contraction force of twisting and braiding configurations in this observation does not conflict with force enhancement in the previous experiments because of different baseline comparisons. The twisting and braiding specimens have shorter initial lengths than their single or straight peers so that they required lower pressure to reach the same elongation (elongation denotes the ratio between accumulative displacement and initial length), resulting in a weaker contraction force. In contrast, if pressure is used as a comparison baseline or input, the twisting and braiding variants will benefit from the twisting angle reduction, enhancing elongation and therefore achieving higher contraction force (Fig. 5.5D).

5.5.5. Multifolded twisting and braiding configurations with a single SMAM

We investigated the performance enhancement of twisting and braiding configurations from another perspective. We utilized only a single SMAM to construct each twisting and braiding
specimen instead of using multiple SMAMs as in the previous comparison. Four specimens were fabricated from the same muscle type (OD 1.49 mm) with different lengths and configurations: single, double-, quadruple- and octuple-twisting so that they have a similar initial length of 35 mm (Fig. 5.6). We then supplied the identical input pressure (equivalent to an input force of 20 N to the syringe plunger) to these specimens and recorded their elongations. Results revealed that the single SMAM could reach 85.7% elongation. This value increased to 91.4% (6.7% enhancement), 100% (16.7% enhancement) and 115.7% (35% enhancement) for double, quadruple and octuple variants, respectively. The results are mean values of ten trials for each configuration.

Figure 5.6. Elongation enhancement by multiple twisting of a single muscle. All four specimens have the initial lengths of 35 mm and were configured from the same fundamental muscles (Table 5.1) with different lengths. Blue arrows: pressurization.
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These results confirmed that the muscle elongation proportionally increases with the number of participating segments or higher twisting turns of the T-SMAM. Similarly, this augmentation mechanism also happens to the B-SMAM. The increment ratio is affected by an increase in both the number of segments and the twisting angle. The specimens with more twisting segments undergo a larger reduction of twisting angles than that of lesser segments, resulting in a larger elongation with the same input pressure. It is also noted that the twisting and braiding configurations of the SMAM share the same applications with the single SMAM but at a higher elongation threshold. As a result, these variants enhance the exertion force with a compact and integrated design, promoting the development of heavy-duty wearable suits or facilitating the miniaturization of medical devices to operate in confined spaces.

5.6. Applications of the SMAM, T-SMAM, and B-SMAM

5.6.1. SMAMs and their associated structures for exoskeleton and wearable assistive devices

Although the proposed SMAM can exert extension force when pressurizing, it belongs to the soft contractible artificial muscle category. The SMAM elongates to store elastic potential energy and subsequently discharges this energy to induce contraction force when depressurizing. The similarity between the SMAM working principle and mechanisms of human biological muscles enables various biomimicking possibilities, where a pair of SMAMs can be employed to manipulate an articulated joint of the human limbs (e.g. knee, ankle, elbow, wrist, or knuckle).

To demonstrate the usefulness of our developed muscles, we fabricated two simple human bone models by 3D printing technique and then we used our muscles to control the joint movement to demonstrate the above concept. The first one was a hand model with an active index finger that has three controlled links and three rotational joints connected via stainless steel pins. We produced a pair of SMAMs (OD3.18 × L100 mm, coated with a layer of Ecoflex-0030, Smooth-On Inc., USA) to mimic the flexor digitorum profundus muscle of the hand. Like the human hand, the integrated SMAMs were not connected directly to the links but via two tendons that run alongside these links and joints (Fig. 5.7A). Despite under-actuation, the two SMAMs could complete the full range of flexion and extension of the index finger (accumulative angle >270°).
Figure 5.7. Bioinspired soft artificial muscles. (A) A pair of single muscles (OD 3.18 × L 100 mm, coated with a layer of Ecoflex-0030; Smooth-On Inc.), which mimic the flexor digitorum profundus muscle of the human hand, could induce 270° bending motion of the 3D-printed index finger. (B) A pair of quadruple-twisting muscles (L 150 mm, made from single muscles OD 1.49 mm), which mimic the human bicep and tricep, could manipulate the 3D-printed elbow joint up to 120° in rotational motion. (C) A pair of double-twisting muscles (length L 130 mm, made from single SMAM with an outer diameter of 3.18 mm), which mimic the human bicep and tricep, could lift a weight of 200 g with 120° in rotational motion.

The second model was an arm-forearm with a rotational elbow manipulated by a pair of quadruple-T-SMAMs (each has a length of L 150 mm, made from a single muscle OD 1.49 × L 600 mm). The two muscles alternate between pressurization and depressurization to mimic the human bicep and tricep, inducing bi-directional rotation of the elbow within a 120° range (Fig. 5.7B). Since the SMAMs exert contraction force, the controlled link will be pulled toward the depressurized muscle. To demonstrate our muscle capability, we replaced the quadruple-twisting muscles with a pair of double-T-SMAMs (each muscle has a length of L 130 mm, made from a single muscle OD 3.18 × L 260 mm) to control the elbow while lifting a weight of 200 g at the hand-end of the forearm (Fig. 5.7C). The double-twisting muscle could generate a torque
of 0.353 Nm (weight of 200 g, arm length of 180 mm) or an equivalent force of 10.1 N (moment arm 35 mm).

It is necessary to mention that we simplified the actual human anatomy by replacing the complex elbow and knuckle joints with the articulated substitutes. These results show that our developed muscles for both SMAMs, T-SMAMs, and B-SMAMs, can be implemented in soft robotic structures and wearable assistive devices for haptic feedback, human augmentation, and rehabilitation.

5.6.2. Growing soft tubular artificial muscles for robotic and medical applications

Growing processes are widely present and occurring every second in nature, where plants and animals evolve over their lifetime. For example, the human esophagus, bones, trachea, and artery can increase their size when the body grows. Likewise, plants also demonstrate an increase in size [42, 43]. There are also many species with tubular organisms that can alternate their body measurements (length and diameter) for either locomotion or prey catching such as bobbit worms (Eunice aphroditois), earthworms, and leeches [44]. Other animals such as snakes or lizards can expand their bodies after swallowing a large-size prey [45, 46]. Inspired by such biological growth, we created new SMAM-based soft tubular muscles that could simultaneously induce longitudinal and radial expansion upon pressurization (Fig. 5.8). The new muscle exhibits the special characteristics of auxetic metamaterial structures, where they have a negative Poisson’s ratio, meaning that they become thicker perpendicular to the axial elongation under applied fluid pressure [47, 48].

To fabricate a soft tubular muscle, we braided eight single SMAMs (OD1.49 × L80 mm) using the hollow round braid technique. The new muscle has an outer diameter of 20 mm, a length of 58 mm, and a twisting angle of 40° (Fig. 5.8). At the interlacing points where any two muscles meet, we used elastic strings to secure these points. It is noted that we used only one hydraulic syringe to supply fluid (distilled water) to all eight SMAMs via a one-to-eight fluid distributor, enabling a miniature size for the transmission tube from the syringe. We supplied input pressure to the soft tubular muscle and this results in simultaneous growth in both axial and radial directions, up to 52% longitudinal elongation ($\varepsilon_l$) and 45% radial expansion ($\varepsilon_r$) (Fig. 5.8B). The underlying mechanism of this growing phenomenon is due to the helical arrangement of each SMAM where the elongation of each individual is divided into longitudinal and radial constituents. The ratio between these two constituents relies on the
twisting angle $\varphi$. For example, the tubular muscle with a twisting angle of $\varphi < 45^\circ$ will have a longitudinal elongation, which is larger than radial expansion and vice versa.

**Figure 5.8.** Bioinspired soft tubular muscle. (A) Inspired by the natural growing process of plants and animals, the hunting and swallowing of some tubular organisms in which their bodies expand longitudinally and radially in a simultaneous manner. (B) Prototype made by hollow round braiding eight single muscles (OD1.49 \times L80 mm). (C) Soft tubular muscle embraces a straight and bending index finger as a smart compression garment (sleeve).
As a proof of concept, we then implemented the soft tubular muscle in several applications. First, the tubular muscle’s radial motion enables its utilization as a compression garment (sleeve), which is one of the most commonly applied methods to treat and prevent a wide variety of musculoskeletal injuries and circulatory conditions [49, 50]. Our tubular muscle at the initial stage could embrace firmly to the human index finger and subsequently release the compression force when receiving input pressure (Fig. 5.8C). Because of its mesh-like structure with high flexibility, the smart compression garment also works well with a bending finger. It means that its mesh eyes can self-adjust their shape to conform to various curved surfaces. This feature allows the use of one long device for multiple links across joints, including arm-elbow-forearm and thigh-knee-lower leg, enabling a smart compression garment for upper limbs and lower limbs. In addition, the soft tubular muscle is scalable due to the scalability of the SMAM’s size and the hollow braid parameters. Therefore, it also permits dimensional customization for specific targeted areas of the human body. The new tubular muscle for compression therapy is expected to potentially provide the normal force to the contact surfaces like other compression garments and a tangential force to the skin while maintaining consistent pressure to reduce pain and inflammation and promote healthy fluid circulation.

Second, we utilize the soft tubular muscle as a support device for medical applications such as medical stents, which can be inserted into the lumen of an anatomic vessel or duct to keep the passageway open or other different purposes ranging from expandable coronary, vascular, and biliary stents, to simple plastic stents used to allow the flow of urine between kidney and bladder [51, 52]. In this paper, our stent-like soft tubular muscles are expected to be used as a new class of soft artificial stents, which can hold arteries open to improve the blood flow or serve as a colon/intestine support device that can prevent collapse while performing endoscopic full-thickness dissection surgeries (Fig. 5.9).

To demonstrate the capability of our soft tubular muscles, we deployed the muscle into a layflat tubing (low-density polyethylene, width 50 mm, thickness 50 µm) and a silicone sleeve (OD19 mm, thickness 0.5 mm, Ecoflex 00-30; Smooth-On, Inc., USA) simulated as a human colon or blood vessel. The tubular muscle plays the role of an expandable soft skeleton providing structural support for a thin and soft outer layer. It could inflate up to 50% $\varepsilon_l$ and 45% $\varepsilon_r$ inside the layflat tubing and drops to 29% $\varepsilon_l$ and 20% $\varepsilon_r$ in the case of the silicone sleeve. These results reveal that the outer layer material properties and thickness greatly affect the tubular muscle growing ratio. Furthermore, the slippery surface of the layflat tubing allows the tubular muscle
to glide inside while growing. This is in contrast with the sticky silicone sleeve where the tubular muscle has to bear the silicone stretching force.

**Figure 5.9.** Soft tubular muscles as hollow support devices. *(Upper panel)* Soft tubular muscle grows inside a silicone sleeve and layflat tubing. *(Lower panel)* Potential application of soft tubular muscle as a medical stent for cardiovascular disease treatment or as a support structure to prevent the collapse of the human colon in endoscopic surgeries.

Finally, we also demonstrated the use of our growing tubular muscle as a soft tubular gripper. Figure 5.10A shows that a tubular muscle (weight of 6 g) could lift a weight of 500 g, which is 83 times heavier than the gripper mass (this already excluded the mass of the miniature
syringe). When approaching the object, we supplied the fluid pressure to expand the muscle diameter to embrace the object. While maintaining the input pressure, we lowered the tubular gripper to further increase the contact friction between the gripper and the target object. Once the gripper conformed to the object, we then released the pressure and moved up the gripper together with the accompanied object (see Supplementary Video S1 for the entire process).

Experimental results revealed that the soft tubular gripper is a promising candidate for object retrieval in confined spaces such as objects lying in a small and long cylindrical tube or foreign objects swallowed inside the human GI tract. Figure 5.10B and C present examples of using the tubular gripper (OD20 \( \times \) L58 mm) to retrieve a painting knife and an “AA” battery, respectively, from a cylindrical tube (ID28 \( \times \) L240 mm). The gripping procedure was similar to the previous lifting demonstration. First, we inserted the tubular gripper (initial phase) into the cylindrical tube and approached the object. Second, we supplied pressure to inflate the gripper until it touches the tube’s inner surface. Third, we performed the swallowing technique by continuously switching between pumping and releasing input pressure while lowering the gripper. These combined maneuvers resulted in the gradual invasion of the gripper to the free space between the tube and the object. Fourth, after embracing/swallowing the object, we released the pressure and lifted the object out of the tube (see Supplementary Video S1 for the swallowing technique).

These results demonstrated that our tubular gripper is also a potential retrieval tool to deal with foreign objects swallowed inside the human GI tract while maintaining sufficient flexibility inside the complex esophagus (Fig. 5.10D). In a retrieval task (e.g. a battery inside the stomach), the gripper with the help of a conventional endoscope will be delivered onsite (stomach) through the upper GI tract. The endoscopist or surgeon will then perform the grasping technique described above to embrace and retrieve the object under endoscopic visualization. The tubular gripper is well suited for retrieving objects comprising many different shapes, especially thin and slender shapes such as bars, beams, and rods. These shapes normally cause difficulty for the current, conventional endoscopic retrieval devices such as retrieval forceps and retrieval nets.
Figure 5.10. Tubular muscle as a tubular gripper. (A) Lifting a weight of 500 g. (B, C) Retrieving a painting knife and an “AA” battery, respectively, from a cylindrical tube (ID28 × L240 mm). (D) Potential use of the tubular gripper to retrieve foreign objects inside the human GI tract. GI, gastrointestinal. White arrows: moving directions of the tubular gripper.
5.7. Discussion and conclusion

We have introduced scalable and flexible SMAMs, created by inserting a soft silicone microtubule into an extension microcoil. This simple fabrication method allows mass production with high repeatability on quality and functionality. Manufacturers can commence large-scale production to suit customers’ muscle specifications or supply spools of long muscles as off-the-shelf products. The SMAM receives fluid pressure to induce longitudinal elongation and store elastic energy simultaneously. Subsequently, it discharges this energy to exert contraction force and return to its initial stage when releasing the pressure. Our SMAM embodiment OD1.49 mm (L75 mm) could achieve 48% elongation when receiving a pressure of 1.05 MPa. It also generated around 0.69 N contraction force when being pressurized to 0.9 MPa. Detailed characterization and performance of our single SMAM such as maximum strain, energy efficiency, durability, and frequency response can be found in our previous work [40].

We exploited the SMAM potential as filaments to configure the more complex and desired structures by twisting and braiding techniques. When supplying the same input pressure (1.05 MPa), compared to multiple single SMAM in a straight arrangement, their twisting and braiding configurations possess higher elongation, at 10.4% and 14.6% enhancement, respectively. In addition to conventional longitudinal elongation upon pressurization, the twisting and braiding variants receive supplemental displacement caused by the decrease of the twisting angle. Consequently, the contraction force of the twisting and braiding variants increased by 11.2%-16.1%. Alongside using multiple single SMAMs, we could establish twisting configurations by multifolded twisting a single SMAM. Experimental demonstrations showed that an octuple-T-SHAM provided 35% more enhancement than a single SMAM with similar initial lengths. The twisting and braiding ability of the SMAM enable elongation and force enhancement without increasing the number of participating muscles, making it suitable for applications with limited spaces or situations requiring compact devices such as endoscopic instruments, soft exoskeleton gloves, or suits.

We have demonstrated the use of our developed SMAM as artificial muscles to mimic the flexor digitorum profundus muscle of the hand to manipulate the entire range of motion of a 3D-printed index finger. We also mimicked the bicep and tricep of a hand model using another two pairs of twisting variants. They could control a 120° motion range of the elbow joint with and without an attached load. This concept of employing a pair of SMAMs to actuate an
articulated joint will be useful in many robotic applications, including surgical robots and exoskeleton devices.

Using the hollow round braid technique, we produced a tubular muscle that could simultaneously grow in the longitudinal and radial directions. Subsequently, we deployed the tubular muscle as a compression sleeve embracing the human index finger to provide massage therapy. Furthermore, it could support hollow cylindrical structures, potentially being used as stents or GI tract support devices. We also implemented the tubular muscle as a tubular gripper that could embrace and lift an object in free space or retrieve objects in a confined space. The gripper also has the potential for foreign object retrieval in the human GI tract.

It is noted that the working principle of our muscle is different from conventional McKibben muscle. In fact, our muscle extends when receiving input pressure and contracts when reducing the pressure. It means our muscle is not used as a pushing actuator but we utilize its contraction force to pull against a load when releasing the pressure. Despite advances, our SMAMs and their associated structures, including T-SHAM, B-SHAM, and SHAMs-based tubular muscle, have several limitations. They require relatively high pressure to function (48% strain at 1.05 MPa) because of the outer spring coil. Since the SHAM is a contractible artificial muscle, the tubular muscle showed good grasping and holding ability but limited expansion force. Therefore, future work should focus on exploring new material combinations to reduce the spring effect, and investigating various braiding techniques or other filament arrangements to further exploit its potential as an artificial muscle.

We believe that a muscle with more strands will produce more elongation enhancement and therefore exert a stronger contraction force. It is definitely possible to increase the number of strands in these configurations to achieve higher force in a trade-off with structural complexity and fabrication difficulty. Theoretically, more strands in a twisting or braiding muscle will increase the twisting angle, translating to a higher enhancement percentage of elongation and force. An optimal number of strands to produce a maximum performance for the muscle can be achieved if a simulation-based optimization technique is used (via Abaqus or COMSOL). However, it is out of the scope of this article. We also hypothesize that the number of participating muscles should stay under a certain threshold, then a higher force demand will be satisfied by increasing the muscle size, which we will carry out in future works.

In our previous work [40], we examined the frequency response of a single muscle up to 20 Hz. Although we have not carried out experiments to examine this effect, we hypothesize that
twisting and braiding configurations will slow down the muscle bandwidth. However, in many practical applications such as prosthesis artificial muscles, hollow support structures, and tubular gripper, the muscle does not require a very high-frequency response. Therefore, the twisting and braiding configurations may achieve a smaller frequency threshold than a single variant, but they probably still meet the required speed for a majority of practical applications.

We also consider adapting the developed SMAM to specific applications including soft exoskeletons, wearable robots, implantable devices, surgical tools, and haptic devices. By outsourcing microtubule and microcoil from leading manufacturers, we strive to produce muscles with 0.3-0.5 mm outer diameter, benefiting many potential medical applications such as microstent, microcatheter, and miniature flexible robotic arms for endoscopic surgery. Furthermore, the current SMAMs and their associated structures do not possess position feedback to monitor muscle elongation. Therefore, we recommend that future work should develop a new integration method of soft conductive materials such as liquid metal alloy or conductive hydrogels [5, 53] into the soft microtubule structure, which subsequently provides either resistive or capacitive sensing. In addition, the muscles still exhibit nonlinear hysteresis profiles, which mainly prevent their use in several applications where precision control of the motion or force is highly desired. Therefore, hysteresis model-based feedforward or nonlinear adaptive controller [54] should be applied to the muscles.

In conclusion, we developed a fluid-driven SMAM that can be used as a single muscle or twisted or braided to enhance performance. Our tubular muscle could handle multiple tasks from a compression sleeve to a hollow support structure, to a tubular gripper to retrieve objects in narrow places. We believe that our muscle construct and its potential application areas will expand knowledge and significantly benefit the robotic community.

References

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Smart textiles made from hydraulic artificial muscles

This chapter explores another versatile aspect of the proposed artificial muscle to create a new type of smart textile. Smart textiles are an emerging research topic in an endeavour to combine the high conformity of traditional passive fabrics with the activeness and controllability of soft artificial muscles. This chapter presents several prototypes of smart textiles created by knitting and weaving fluid-driven artificial muscle fibres. The prototypes are highly flexible and capable of producing multimodal motion and shape-shifting capabilities such as elongation, area expansion, radial expansion and bending motion. Smart textile characteristics are experimentally measured. Analytical models to describe the relationship between elongation and force of knitted and woven sheets are developed and validated. Furthermore, this chapter also introduces the concept of bio-inspired shape-morphing structures made by sticking artificial muscles to conventional fabrics. This method allows passive flat sheets to be easily reconfigured into active three-dimensional structures. This chapter serves as a proof-of-concept study that paves the way for further exploration of smart wearable devices and biomimetic soft robots.

This work has been published:

6.1. Abstract

The marriage of textiles with artificial muscles to create smart textiles is attracting great attention from the scientific community and industry. Smart textiles offer many benefits including adaptive comfort and high conformity to objects while providing active actuation for desired motion and force. This paper introduces a new class of programmable smart textiles created from different methods of knitting, weaving, and sticking fluid-driven artificial muscle fibers. Mathematical models are developed to describe the elongation-force relationship of the knitting and weaving textile sheets, followed by experiments to validate the model effectiveness. The new smart textiles are highly flexible, conformable, and mechanically programmable, enabling multimodal motions and shape-shifting abilities for use in broader applications. Different prototypes of the smart textiles are created with experimental validations including various shape-changing instances such as elongation (up to 65%), area expansion (108%), radial expansion (25%), and bending motion. The concept of reconfiguring passive conventional fabrics into active structures for bio-inspired shape-morphing structures is also explored. The proposed smart textiles are expected to contribute to the progression of smart wearable devices, haptic systems, bio-inspired soft robotics, and wearable electronics.

6.2. Introduction

Rigid robots are effective when working in structured environments but encounter problems dealing with unknown contexts of changing environments, thus restricting their applications for searching or exploration. Nature always amazes us with numerous smart strategies to handle external factors and versatilities. For example, the tendrils of climbing plants perform multimodal motions such as bending and spiral twisting to explore the unknown environment to find suitable supports [1]. The Venus flytrap (Dionaea muscipula) is equipped with sensitive hairs on its leaves that upon trigger will snap shut to catch prey [2]. Shape-morphing or shape-shifting bodies from two-dimensional (2D) surfaces to three-dimensional (3D) shapes which mimic biological structures have become interesting research topics in recent years [3, 4]. These soft robotic configurations alter their shapes to adapt to versatile environments, provide multimodal motions, and exert force to generate mechanical work. Their reaches have been expanded to a wide range of robotic applications, including deployable structures [5], reconfigurable and self-folding robots [6, 7], biomedical devices [8], locomotion [9, 10], and stretchable electronics [11].
Chapter 6. Smart textiles made from hydraulic artificial muscles

Many studies have been conducted to develop a programmable planar sheet that transforms into a complex 3D structure upon activation [3]. A simple idea to generate shape-shifting structures is to combine layers of different materials that yield bending and wrinkle motions when triggered by stimuli [12, 13]. This concept has been implemented by Janbaz et al. [14] and Lee et al. [15] to produce thermally-responsive multimodal shape-shifting robots. Origami-based structures incorporated with stimuli-responsive elements have been exploited to create complex 3D structures [16-18]. Inspired by the morphogenesis of biological structures, Emmanuel et al. created shape-morphing elastomers by arranging airways inside a rubber surface, which transform into complex arbitrary 3D shapes upon pressurization [19].

Integrating textiles or fabrics into shape-morphing soft robots is another emerging conceptual design that attracts intensive interest. Textiles are soft and flexible materials made from yarns by interlacing techniques such as knitting, weaving, braiding, or knotting. Textiles have amazing characteristics, including flexibility, conformability, stretchability, and breathability, making them extremely popular in every aspect of life from clothing to medical applications [20]. There are three broad approaches to incorporating textiles into robotics [21]. The very first approach is to use textiles as passive substrates or the base to house other components. In this circumstance, passive textiles provide a comfortable fit to the users while carrying rigid components (motors, sensors, power suppliers). Most soft wearable robots or soft exoskeletons belong to this approach. For example, soft wearable exoskeletons for walking assistance [22] and elbow joint assistance [23-25], a soft wearable glove for hand and finger assistance [26], and bio-inspired soft robots [27].

The second approach is to use textiles as passive and constrained components of soft robotic devices. Textile-based actuators fall into this category where textiles are normally constructed as external containers to constrain inner soft tubes or bladders which form soft fibre-reinforced actuators. Upon pressurization by an external pneumatic or hydraulic source, these soft actuators exert shape-change, either elongating, bending, or twisting, according to their initial compositions and configurations. For example, Thalman et al. introduced an ankle-foot orthosis exosuit made from a series of fabric pouches to assist in plantar flexion for gait rehabilitation [28]. Textile layers with different stretchability could be combined to produce anisotropic motions [29]. OmniSkins – soft robotic skins made of various soft actuators and substrate materials could allow the transformation of passive objects into multifunctional active robots that could perform multimodal locomotion and deformations for various applications [30]. Zhu et al. developed fluidic fabric muscle sheets that produced elongating, bending, and
various shape-shifting motions [31]. Buckner et al. integrated functional fibers into conventional fabrics to create robotic fabrics that had multiple functions such as actuation, sensing, and variable stiffness [32]. Other approaches in this category can be found in these works [21, 33-35].

The latest approach which leverages excellent textile properties in the soft robotic field is to use active yarns or stimuli-responsive filaments to construct smart textiles by using traditional textile-making approaches such as braiding, knitting, and weaving techniques [21, 36, 37]. Depending on material compositions, the active yarns induce shape-change when triggered by electrical, thermal, or pressure inputs, resulting in textile deformation. In this approach, the shape-change of textiles happens at the interior level (yarns) rather than the exterior level when integrating conventional textiles into soft robotic systems. Therefore, smart textiles offer great controllability in terms of multimodal motion, programmable shape-shifting, stretchability, and stiffness tuneability. For example, shape memory alloys (SMAs) and shape memory polymers (SMPs) could be incorporated into fabrics to actively control their shapes by thermal excitation such as a roll-up hemline [38], wrinkle recovery [36, 39], tactile and haptic feedback [40, 41], and self-fitting wearables [42]. However, the use of thermal heating and cooling resulted in slow response, complex cooling, and control. Recently, Hiramitsu et al. implemented thin McKibben muscles [43, 44] (a type of pneumatic artificial muscles) as warps to create multiple forms of active textiles by changing woven structures [45]. Although this approach offers high force, it was limited at the expansion rate (<50%) and could not achieve a small size (<0.9 mm in diameter) due to the nature of the McKibben muscle. In addition, it found challenges to form smart textile patterns from the knitting method, which requires a sharp bending angle. To form a larger array of smart textiles, Maziz et al. developed electroactive textiles for wearable devices by knitting and weaving electrically responsive polymer yarns [46].

Recent years have witnessed the emergence of a new class of thermally-responsive artificial muscles constructed from highly twisted inexpensive polymer fibers [47, 48]. These fibers are commercially available and easy to incorporate into machine-knitted or -woven to produce affordable smart garments. Despite advances, these new thermal smart textiles are limited with slow response time due to the requirement of heating and cooling (e.g. thermally controlled textiles) or difficulties to fabricate complex knitting and weaving patterns that can be programmed to form desired deformations and motions such as radial expansion, shape transformation from 2D to 3D or bidirectional expansion as we propose here.
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To overcome these aforementioned challenges, this paper introduces a new class of fluid-driven smart textiles formed from our recently soft artificial muscle fibers (AMFs) [49-51]. The AMFs are highly flexible, scalable with sizes that can be scaled down to 0.8 mm in diameter, long length (at least 5000 mm) which offers a high-aspect ratio (length-per-diameter) as well as high elongation (at least 245%), high energy efficiency, and fast response at least 20 Hz. To create smart textiles, we employ AMFs as active yarns to form 2D active muscle sheets by knitting and weaving techniques. We quantitatively examine the expansion ratio and contraction force of these smart textiles with respect to applied input fluid volume and pressure. Analytical models have been developed to establish the elongation-force relationship of knitted and woven sheets. We also introduce several techniques to mechanically program the smart textiles to achieve multimodal motion including bidirectional elongation, bending, radial expansion, and growing ability from 2D to 3D. To demonstrate the capability of our approaches, we also integrate the AMFs into commercial fabrics or textiles to reconfigure them from passive to active structures that can induce different deformations. We also demonstrate this concept via several experimental testbeds including programmable bending filaments to achieve desired letters and bio-inspired shape-morphing structures in the form of objects such as a butterfly, four-legged structures, and a flower with complex motions.

6.3. Results

6.3.1. Smart textile composition and configurations

A textile is a flexible two-dimensional structure created by interlocking one-dimensional filaments such as yarns, threads, and fibers. Textiles are one of humankind’s oldest technologies, being widely used in every aspect of life thanks to their comfort, adaptability, breathability, appearance, and protection. Smart textiles (also known as smart garments or robotic fabrics) have increasingly been used in research studies because of their vast potential in robotic applications [20, 52]. Smart textiles are expected to enhance human experiences to interact with soft bodies, opening a paradigm shift in the field where a thin and flexible piece of fabric can be controlled in motion and force to perform specific tasks. In this work, we explore two methods to produce smart textiles based on our recent AMFs: [49] (1) using the AMFs as active yarns to construct smart textiles by traditional textile-making techniques; (2) directly sticking the AMFs into conventional fabrics to induce desired motion and deformation.
The AMFs consist of an inner silicone tube to receive hydraulic power and an outer helical coil to constrain its radial expansion. Therefore, AMFs produce longitudinal elongation upon pressurization and subsequently exert contraction force when releasing the pressure to return to the initial length. They possess similar characteristics as traditional fibers including flexibility, small diameter, and long length. However, the AMFs surpass their traditional counterparts by being active and controllable in terms of motion and force. Motivated by recent fast-paced developments in the smart textile field, we introduce here four main approaches to producing smart textiles by implementing the AMFs into long-established fabric-making techniques (Fig. 6.1).

**Figure 6.1.** Different approaches to create smart textiles from artificial muscle fibers.

The first approach is knitting. We utilize the weft knitting technique to produce an active knitting sheet that can expand in one direction upon hydraulic pressurization. Knitting sheets are highly elastic and highly stretchable but more prone to unravelling than weaving sheets. One AMF can form a single course or an entire knitting sheet depending on control approaches. Besides the planar sheet, tubular knitting patterns are applicable for AMF to make hollow
structures. The second approach is weaving where we make use of two AMFs as warp and weft to form a rectangular weaving sheet that can exert independently bidirectional expansion. Weaving sheets offer a greater control degree (two directions) compared to knitting sheets. We also weave a single AMF with conventional yarns to produce a simpler weaving sheet that can only expand in one direction. The third approach is radial expansion—a variant of the weaving technique where instead of being rectangularly arranged, the AMF is aligned in a spiral shape with yarns providing radial constraint. In this case, the weaving sheet will expand radially when receiving input pressure. The fourth approach is to stick the AMFs to a passive fabric sheet to create bending motions in desired directions. We reconfigure a passive fabric sheet into an active one by routing an AMF around its boundary. This programmable characteristic of the AMF opens numerous possibilities for bio-inspired shape-morphing soft structures where we can turn passive objects into active ones. This approach is simple, easy, and fast but may compromise the prototypes’ durability. Readers can refer to other approaches from the literature which detailed the pros and cons of the performance of each type of textile [21, 33-35].

6.3.2. AMF configurations to produce smart textiles

Most filaments or yarns which produce conventional textiles comprise a passive structure. In this work, we take advantage of our previously developed AMFs that can be made meters in length with sub-millimeter diameters, replacing the conventional filaments of the passive textiles by the AFMs to form smart and active textiles for broader applications. The following sections describe detailed fabrication methods to produce smart textile prototypes as well as introduce their basic characteristics and behaviors.

**Knitting sheet**

We manually fabricated a knitting sheet from three AMFs using the weft knitting technique (Fig. 6.2A). Material selections and detailed specifications of AMFs and prototypes can be found in the “Methods” section. Each AMF followed a meandering path (also known as a course) that formed symmetric loops. The loops of each course were secured with the loops of the courses just above and below them. Loops of the same column that were perpendicular to the course were grouped into a wale. Our knitting prototype consisted of three courses with seven loops for each course (or seven wales). The loops of the top and bottom courses were not secured so that we could connect them to corresponding metal rods. The knitting prototype was more prone to unravel than conventional knitting fabric because of the higher stiffness of
AMFs compared to conventional yarns. Therefore, we constrained loops of adjacent courses by a thin elastic string.

Although the meandering loops of the knitting sheet could be stretched in different directions, our knitting prototype primarily expanded in the wale direction upon pressurization owing to the constraint in the course direction. The elongation of each AMF contributed to the overall area expansion of the knitting sheet. Depending on specific requirements, we could control three AMFs independently from three different fluid sources (Fig. 6.2A) or simultaneously from a single fluid source via a 1-to-3 fluid distributor. Figure 6.2A shows one instance of the knitting prototype that expanded 35% of its initial area when pressurizing (1.2 MPa) three AMFs simultaneously. It is noted that the AMF achieved a high elongation of at least 250% of its initial length [49] and therefore the expansion ratio of the knitting sheet can be made higher than that of the current version.

**Weaving sheet**

We also created a bidirectional weaving sheet formed from two AMFs by using the plain weaving technique (Fig. 6.2B). The warp and weft AMFs were interlaced at right angles to form a simple criss-cross pattern. Our weaving prototype was classified as balanced plain weaves because both warp and weft were made of filaments of the same size (detailed specifications are shown in the “Methods” section). Unlike conventional yarns which are capable of sharp folds, the used AMFs require a certain bending radius when turning back for another line of the weaving pattern. As a result, the weaving sheet made from the AMFs is less dense compared to conventional weaving textiles. The minimum bending radius of AMFs type S (OD 1.49 mm) is 1.5 mm. As an illustration, our weaving prototype presented in this paper had a pattern of 7 × 7 lines where each cross was stabilized by a knot made of a thin elastic string. A higher number of lines can be achieved using the same weaving technique.

The weaving sheet expanded its area toward the warp or weft direction when the corresponding AMF received fluid pressure. Therefore, we controlled the weaving sheet dimension (length and width) by independently varying the amplitude of applied input pressure to the two AMFs. Figure 6.2B shows the weaving prototype that expanded 44% of its initial area when pressurizing (1.3 MPa) one AMF at a time. An area expansion of 108% was achieved when simultaneously pressurizing the two AMFs.

We also fabricated a unidirectional weaving sheet formed from a single AMF as warp and acrylic yarns as weft (Fig. 6.2C). The AMF was arranged in a seven-line zigzag whereas the
yarns interlaced these AMF lines to form a rectangular textile sheet. This weaving prototype was denser than the one in Fig. 6.2B thanks to the soft acrylic yarns that easily filled the entire sheet. Since we used only one AMF as the warp, the weaving sheet can only expand toward the warp direction upon pressurization. Figure 6.2C shows one instance of the weaving prototype that expanded 65% of its initial area when pressurizing (1.3 MPa). In addition, this weaving sheet (which weighed 2.6 g) could lift a load of 500 g which is 192 times heavier than its mass.

Figure 6.2. Different prototypes of smart textiles are achieved by varying AMF configurations. (A) Knitting sheet made from three AMFs. (B) Bidirectional weaving sheet made from two AMFs. (C) A unidirectional weaving sheet made from one AMF and acrylic yarns could lift a 500 g load, which is 192 times heavier than its mass (2.6 g). (D) Radial expansion structure made from a single AMF and cotton yarns as radial constraints. Detailed specifications can be found in the “Methods” section.
Radial expansion with circular weaving sheet

Instead of arranging an AMF in a zigzag layout to create a rectangular weaving sheet, we made a planar spiral shape of an AMF and then constrained it radially with cotton yarns to create a circular weaving sheet (Fig. 6.2D). The high stiffness of the AMF restrained it from filling the centermost area of the circular sheet. However, this filling can be done with elastic yarns or stretchable fabrics. When receiving hydraulic pressure, the AMF transformed its longitudinal elongation into the radial expansion of the sheet. It is also worth noting that both the outer and inner diameters of the spiral shape are expanded, due to the radial constraint of the yarns. Figure 6.2D shows the shape of the circular sheet that expanded 25% of its initial area under applied hydraulic pressure of 1 MPa.

Fabric reconfiguration

We introduce here the second method to create smart textiles where we stick an AMF to a piece of planar fabric to reconfigure it from a passive to an active, controllable structure. The design concept of a bending actuator is illustrated in Fig. 6.3A where an AMF is folded at its midpoint and stuck to a strip of non-stretchable fabric (cotton muslin fabric) using double-sided tape as a bonding element. Upon pressurization, the top part of the AMF elongates freely whereas the bottom part is constrained by the tape and fabric, resulting in a bending motion of the strip toward the fabric side. We can deactivate any segment at any location of the bending actuator by simply putting a piece of tape on top of it. The deactivated segments are unable to exert any motion and become passive segments.

We fabricated several bending actuator prototypes with different lengths and hydraulically pressurized them to generate the bending motion (Fig. 6.3B). It is noted that the AMF can be arranged in a straight line or folded to form multiple lines before sticking to the fabric to create a bending actuator with the corresponding number of lines. We also reconfigured a passive fabric sheet into an active four-legged structure (Fig. 6.3C) where we routed the boundaries of rectangular non-stretchable fabric (cotton muslin fabric) with an AMF. The AMF was stuck to the fabric with the help of a piece of double-sided tape. The middle segments of each edge were taped to become passive, leaving four corners active. A top cover made of stretchable fabric (polyester) was optional. The four corners of the fabric were bent down (looked like legs) when pressurized.
Figure 6.3. Fabric reconfiguration by sticking AMFs to conventional fabrics. (A) Design concept of bending actuators made by sticking a folding AMF to a non-stretchable fabric. (B) Bending actuator prototype. (C) Reconfiguring a rectangular fabric into an active four-legged robot. Non-stretchable fabric: plain-woven cotton muslin; stretchable fabric: polyester. Detailed specifications can be found in the “Methods” section.
6.3.3. Characteristics of smart textiles

*Input pressure versus output elongation and force*

We built a testing platform to quantitatively examine the characteristics of the developed smart textiles (see the “Methods” section and Supplementary Fig. 6.S1). Since all specimens were made of AMFs, the overall trend of experimental results (Fig. 6.4) agrees with the AMF’s fundamental characteristic where the input pressure has a proportional relationship with output elongation and a reverse proportional relationship with contraction force. However, these smart textiles have their own distinctive features that represent their particular configurations.

Each AMF of the weaving sheet received 1 MPa input pressure to generate approximately 30% elongation (Fig. 6.4A). We selected this threshold for the whole experiment for several reasons: (1) to create a substantial elongation (around 30%) to highlight their hysteresis profile, (2) to prevent unexpected damage or malfunction resulting from cyclic motion and reusable prototypes for different experiments under high fluid pressure. A dead zone was clearly visible, where the weaving sheet remained static until input pressure reached 0.3 MPa. The pressure-elongation hysteresis chart showed a wide gap between the pressurizing and releasing phases, indicating a significant energy loss when the weaving sheet changes its motion from expansion to contraction. (Fig. 6.4A). The weaving sheet could exert a contraction force of 5.6 N after receiving 1 MPa input pressure (Fig. 6.4B). The pressure-force hysteresis chart also showed that the releasing curve almost overlapped the pressurizing curve. The weaving sheet’s area expansion relies on the pressure amplitudes supplied to each of the two AMFs, which are shown in the three-dimensional surface plot (Fig. 6.4C). Experiments also revealed that the weaving sheet could generate an area expansion of 66% when its warp and weft AMFs simultaneously received 1 MPa hydraulic pressure.

Experimental results of the knitting sheet have a similar pattern to that of the weaving sheet, including the wide hysteresis gap in the pressure-elongation chart and the overlapping curves in the pressure-force relation. The knitting sheet generated 30% elongation and subsequently exerted a contraction force of 9 N when receiving 1 MPa input pressure (Fig. 6.4D,E).

In the case of the circular weaving sheet, there was an expansion from its initial area to 25% after receiving 1 MPa fluid pressure (Fig. 6.4F). There was a substantial dead zone of input pressure until 0.7 MPa before the specimen started to expand. This large dead zone is expected because the specimen was made of a larger AMF that required a higher pressure to overcome its initial tension. Figure 6.4F also showed that the releasing curves almost overlapped the
pressurizing curves, denoting inconsiderable energy loss when switching the circular sheet motions.

Experimental results for three bending actuators (fabric reconfiguration) showed that their hysteresis profiles share similar patterns (Fig. 6.4G), in which they underwent a dead zone of input pressure until 0.2 MPa before rising. We supplied the same volume of fluid (0.035 mL) to the three bending actuators (L20, L30, and L50 mm). However, each actuator experienced different pressure peaks and generated different bending angles. The L20 and L30 mm actuators experienced 0.72 and 0.67 MPa input pressure to reach a bending angle of 167° and 194°, respectively. The longest bending actuator (L50 mm) experienced 0.61 MPa pressure to achieve the largest bending angle of 236°. The pressure-angle hysteresis chart also showed a relatively wide gap between the pressurizing and releasing curves of all three bending actuators.

The relationship between input volume and output characteristics (elongation, force, area expansion, bending angle) of the above smart textile configurations can be found in Supplementary Fig. 6.S2.

Figure 6.4. Characteristics of smart textile configurations. (A, B) Hysteresis profiles of input pressure and output elongation and force of the weaving sheet. (C) Area expansion of the weaving sheet. (D, E) Relationship between input pressure and output elongation and force of the knitting sheet. (F) Area expansion of the radial expansion structure. (G) Bending angles of three different lengths of bending actuators.
Chapter 6. Smart textiles made from hydraulic artificial muscles

Elongation-force relationship of knitting and weaving sheets
Experimental results in the previous section clearly demonstrated the proportional relation
between the applied input pressure and the output elongation of specimens made of AMFs. The
more pressure the AMF receives, the more elongation it generates and the more elastic energy
it accumulates. Consequently, the more contraction force it exerts. The results also revealed
that the specimens reached their maximum contraction force when input pressure was
completely withdrawn. This section aims to establish the direct relationship between elongation
and maximum contraction force of knitting and weaving sheets by both analytical models and
experimental validation.
The maximum contraction force Fout of a single AMF (when input pressure P = 0) has been
presented in reference [49], which is reintroduced as follows:
𝐹𝑜𝑢𝑡 = 𝛼𝐸𝐴0 (1 −

1
) + 𝑘𝑥
1 + 𝑥 ⁄ 𝑙𝑖

(1)

where , E, A0 represents the stretch ratio, Young’s modulus, and cross-sectional area of the
silicone tube, respectively; k is the stiffness coefficient of the helical coil; x and li are the
displacement and initial length of the AMF, respectively.
Adapting Eq. (1) to the case of knitting and weaving sheets (Fig. 6.5A,B). The contraction
force of a knitting sheet Fkv and a weaving sheet Fwh are expressed by Eqs. (2) and (3),
respectively.
𝐹𝑘𝑣 = 2𝑚𝑘 [𝛼𝐸𝐴0 (1 −

1
𝑞𝑥 ,
𝜀𝑘𝑣 ≤ 2%
) + { 𝑘𝑣
] cos 𝜑𝑝
𝑘𝑥
+
𝐹
,
1 + 𝑥𝑘𝑣 ⁄𝑙𝑘𝑖
𝑘𝑣
0 𝜀𝑘𝑣 > 2%

(2)

Continuity condition: 𝑞 = 𝑘 + 𝐹0 ⁄𝑥𝑘𝑣 , 𝜀𝑘𝑣 = 2%
𝐹𝑤ℎ = 𝑚ℎ [𝛼𝐸𝐴0 (1 −

1
𝑞𝑥 ,
𝜀𝑤ℎ ≤ 2%
) + { 𝑤ℎ
] cos 𝜃ℎ𝑝
𝑘𝑥𝑤ℎ + 𝐹0 , 𝜀𝑤ℎ > 2%
1 + 𝑥𝑤ℎ ⁄𝑙ℎ𝑖

(3)

Continuity condition: 𝑞 = 𝑘 + 𝐹0 ⁄𝑥𝑤ℎ , 𝜀𝑤ℎ = 2%
where mk is the number of wales and p is the looping angle at the pressurizing phase of the
knitting sheet (Fig. 6.5A); mh is the number of lines and hp is the interlocking angle at the
pressurizing phase of the weaving sheet (Fig. 6.5B); kv and wh are the strain of the knitting
sheet and the weaving sheet, respectively; F0 is the initial tension of the helical coil. Detailed
derivations of Eqs. (2) and (3) can be found in the Supporting Information.
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To validate the developed models, we conducted elongation-force experiments using the knitting specimen in Fig. 6.2A and the weaving specimen in Fig. 6.2B. The contraction force was collected at each blocked elongation from 0 to 50% with a 5% increment. Mean values and standard deviation of five trials were plotted in Fig. 6.5C (for knitting) and Fig. 6.5D (for weaving). The analytical model curve was governed by Eqs. (2) and (3) with parameters as shown in Table 6.1. The results show that the analytical models closely followed experimental data for the entire elongation range with a root mean square error (RMSE) of 0.34 N for knitting, 0.21 N for weaving AMF H (horizontal direction), and 0.17 N for weaving AMF V (vertical direction).

Figure 6.5. Analytical models to establish the elongation-force relationship. (A, B) Analytical model illustration for knitting and weaving sheets, respectively. (C, D) Comparison between analytical models and experimental data for knitting and weaving sheets, respectively. RMSE: root mean square error.
Table 6.1. Parameters of analytical models. $k$ and $F_0$ at length 50 mm, $E$ at 100% strain.

<table>
<thead>
<tr>
<th>AMF</th>
<th>Helical coil</th>
<th>Silicone tube</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$d$ 1.49 (mm)</td>
<td>$\alpha$ 0.9</td>
</tr>
<tr>
<td></td>
<td>$k$ 0.011 (N/mm)</td>
<td>$E$ 1.648 (MPa)</td>
</tr>
<tr>
<td></td>
<td>$F_0$ 0.126 (N)</td>
<td>$A_0$ 0.8026 (mm$^2$)</td>
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<table>
<thead>
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<th>Configuration</th>
<th>Knitting</th>
<th>Weaving</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>$n$ 3</td>
<td>$m_h$ 7</td>
</tr>
<tr>
<td></td>
<td>$m_k$ 7</td>
<td>$m_v$ 7</td>
</tr>
<tr>
<td>$h_{ki}$</td>
<td>54 (mm)</td>
<td>$h_{wi}$ 52 (mm)</td>
</tr>
<tr>
<td>$v_{ki}$</td>
<td>52 (mm)</td>
<td>$v_{wi}$ 52 (mm)</td>
</tr>
<tr>
<td>$\varepsilon_{kv}$</td>
<td>0 – 50 (%)</td>
<td>$\varepsilon_{wh}$ 0 – 50 (%)</td>
</tr>
</tbody>
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6.3.4. Shape-programmability of smart textiles

In addition to the basic motions, the proposed smart textiles can be mechanically programmed to provide more complex motions such as s-shape bending, radial compression, and shape-shifting from 2D to 3D. We introduce here several techniques to program a planar smart textile into desired structures.

**Unidirectional weaving sheet**

Alongside area expansion in a straight direction, unidirectional weaving sheets can be mechanically programmed to produce multimodal motion (Fig. 6.6A). We reconfigured the weaving sheet elongation into bending motion by constraining its one face (either top or bottom) using sewing thread. The sheet tends to bend toward the constraining face upon pressurization. Figure 6.6A shows two instances of the weaving sheet, where it transformed into an s-shape when constraining one half at the top face and the other half at the bottom face. Alternatively, when constraining only one entire face a loop bending motion could be generated. The unidirectional weaving sheet can also be implemented as a compression sleeve by uniting its two ends to form a tubular structure (Fig. 6.6B). The sleeve could enclose the human index finger and provide compression force, which is massage therapy to relieve pain.
or improve blood circulation. It can be scaled up to fit other body parts such as arms, thighs, and legs.

Figure 6.6. Capabilities of unidirectional weaving sheets. (A) Shape programmability by sewing thread to produce shape-shifting structures. (B) Compression sleeve for a finger. (C) Another weaving sheet embodiment and its implementation as a forearm compression sleeve. (D) Another compression sleeve prototype made of an AMF type M, acrylic yarns, and a Velcro strap. Detailed specifications can be found in the “Methods” section.
Figure 6.6C presents another embodiment of the unidirectional weaving sheets made of a single AMF and cotton yarns. The sheet could generate 45% area expansion (at 1.2 MPa) or induce a looping motion upon pressurization. We also implemented the sheet to create a compression sleeve for the forearm by attaching a magnetic strap to the sheet’s ends. Another compression sleeve prototype for the forearm is shown in Fig. 6.6D where the unidirectional weaving sheet was made of an AMF type M (see the “Methods” section) and acrylic yarns to generate a stronger compression force. We equipped the sheet’s ends with Velcro straps to facilitate attachment and adapt to various arm sizes.

Bidirectional weaving sheet

The constraining technique to transform straight elongation into bending motion is also applicable for the bidirectional weaving sheet. We interlaced cotton yarns to one face of the weaving sheet in both warp and weft directions to impede its expansion (Fig. 6.7A). Therefore, when two AMFs independently received hydraulic pressure, the sheet exerted a bidirectional bending motion, forming arbitrary 3D structures. On the other approach, we used inextensible yarns to constrain one direction of the bidirectional weaving sheet (Fig. 6.7B). As a result, the sheet could generate independent bending and elongating motion when pressurizing the corresponding AMFs. Figure 6.7B showed one instance in which the bidirectional weaving sheet was controlled to envelop two-thirds of the human finger by bending motion and then extend its length to cover the remaining by elongating motion. The bidirectional motion of the sheet may benefit fashion design or smart garment development.

Knitting sheet

We joined two adjacent loops of the top and bottom courses of the knitting sheet by sewing thread to secure it from unraveling (Fig. 6.7C). Consequently, the knitting sheet is entirely flexible and highly conformable to various surface curvatures such as skin surfaces of the human hand and arm. We also created a tubular structure (sleeve) by uniting two ends of the knitting sheet in the course direction. The sleeve nicely embraced the human index finger (Fig. 6.7C). The meandering features of the knitting sheet offer great conformability and deformability, facilitating its use in smart garments (gloves, compression sleeves) that provide the wearers with comfort (by adaptability) and therapeutic effect (by compression force).
Figure 6.7. Capabilities of a bidirectional weaving sheet, knitting sheet, and radial expansion structure. (A) Bidirectionally constraining a bidirectional weaving sheet to produce bidirectional bending. (B) Unidirectionally constraining a bidirectional weaving sheet to produce bending and elongating. (C) Highly conformable knitting sheet that could adapt to various surface curvatures or even form a tubular structure. (D) Constraining the centerline of a radial expansion structure to form a hyperbolic paraboloid shape (a potato chip).

**Radial expansion**

Besides 2D radial expansion in multiple directions, the circular weaving sheet can also be programmed to form a 3D structure. We constrained the centerline of the circular weaving sheet with acrylic yarns to disrupt its uniform radial expansion. As a result, the initial planar shape of the circular weaving sheet was transformed into a hyperbolic paraboloid-like shape (or a potato chip) upon pressurization (Fig. 6.7D). This shape-shifting capability can be implemented as a lifting mechanism, optical lens, legs for locomotion robots, or maybe useful for fashion design and bio-mimicking robots.
Figure 6.8. Fabric reconfiguration to produce shape-morphing structures. (A) Sticking an AMF to passive fabric sheet’s boundaries to turn it into a controllable four-legged structure. (B-D) Another two examples of fabric reconfiguration that turn the passive fabric butterfly and flower into active ones. Non-stretchable fabric: plain-woven cotton muslin.
Fabric reconfiguration

We have developed a simple technique to create a bending actuator – by sticking an AMF to a strip of non-stretchable fabric (Fig. 6.3). We leveraged this concept to create shape-programmable filaments, in which we can strategically allocate multiple active and passive segments in a single AMF to produce the desired shape. We fabricated and programmed four active filaments that could transform their shapes from straight lines into letters (UNSW) upon pressurization (Supplementary Fig. 6.S4). This simple technique enables the shape-shifting capability of AMFs to turn 1D lines into 2D shapes and possibly 3D structures.

In a similar approach, we utilized a single AMF to reconfigure a piece of passive, conventional fabric into an active four-legged structure (Fig. 6.8A). The routing and programming concept was similar to the one in Fig. 6.3C. However, we began with a fabric shaped in a four-legged pattern (turtle-like shape, cotton muslin fabric) instead of utilizing a rectangular sheet. As a result, the legs were longer and could raise the structure higher. The structure’s height gradually increased when pressurizing until its legs were perpendicular to the ground. If input pressure kept rising, the legs would bend inward, lowering the structure’s height. The four-legged structure can exert locomotion if its feet are equipped with unidirectional patterns or using multiple AMFs with locomotion operation strategies. Locomotion soft robots are needed in various tasks, including rescue missions from bushfires, collapsed buildings, or hazardous environments, and drug delivery robots for medical applications.

We also leverage the simplicity and versatility of this fabric reconfiguration technique by introducing two other bio-inspired shape-morphing structures (Fig. 6.8B–D). These shape-morphing structures were reconfigured from passive fabric sheets into active and controllable ones with the help of routing AMFs. Inspired by the Monarch butterfly, we fabricated a butterfly morphing structure using a piece of butterfly-shaped fabric (cotton muslin fabric) and a long AMF stuck under its wings. The wings will bend upward when pressurizing the AMF. Like the Monarch butterfly, both the left and right wings of the butterfly robot were flapping in the same manner because they were controlled by a single AMF. The butterfly was flapping for demonstration only. It could not fly like the Smart Bird (Festo Corp., USA). We also fabricated a fabric flower (Fig. 6.8D) which had two layers of petals, five petals for each layer. We stuck a single AMF beneath each layer following the outer boundaries of the petals. Initially, the flower was fully blooming, in which all petals were entirely open. Upon pressurization, the AMF caused the bending motion of petals, making them close. Two AMFs
controlled the motion of two layers independently while five petals in the same layer were curving at the same time.

Thanks to the fast response characteristic of AMFs, these shape-morphing structures could perform at a comparative speed (2 Hz for the four-legged structure and 1 Hz for the butterfly and the flower) and high durability or repeatability (over 1000 cycles).

6.4. Discussion and conclusion

The textile-based robotic field is undergoing a paradigm shift from using textiles as passive substrates to programmable smart textiles whose compositions consist of stimuli-responsive yarns. Our recently developed soft actuators (or AMFs) which resemble conventional passive yarns in terms of flexibility and high length-per-diameter ratio can elongate back and forth under hydraulic pressure, being excellent candidate elements to fabricate smart and active textiles. We have presented two approaches to creating AMF-based smart textiles: (1) knitting and weaving AMFs; and (2) sticking AMFs to conventional passive fabrics.

In the first approach, we adapted traditional knitting and weaving techniques to construct an AMFs-based knitting sheet, a bidirectional weaving sheet, a unidirectional weaving sheet, and a circular weaving sheet. In their current experimental configurations, these smart textiles could generate an area expansion of 35%, 108%, 65%, and 25%, respectively. Our results are comparable with those in the literature, for example, the knitted textiles made of electroactive polymers achieved less than 5% strain [46], the active textiles made of McKibben muscles exerted a maximum contraction ratio of 7.1% [45], the knitted SMA wrist sleeve achieved 36% contraction [42]. Generally, the AMF (which is fluid-driven) has a faster response time compared to thermally-responsive filaments. In addition, the elongation rate of these smart textiles can reach a higher value of at least 250% which is the elongation limit of the AMF as demonstrated in our recent work [49].

A hysteresis gap occurred in the pressure-elongation hysteresis profiles, being an inherent characteristic of the AMFs. It reflects energy loss when switching from the pressurizing phase to the releasing phase. Experimental data revealed that the hysteresis gaps of knitting and weaving configurations were significantly wider than that of a single AMF [49]. It means that the hysteresis of AMFs is an accumulative attribute. The dead zones found in hysteresis charts represent the initial tension of each particular configuration. Input energy (pressure) is required
to overcome these thresholds in order to change the configuration status from static to expanding. The initial tension of AMFs is also an accumulative attribute and is heavily correlated with the elastic properties of the constituent helical coil and silicone tube. When switching the AMF operation mode from pressurizing to releasing, energy loss was in the form of heat. However, since most of our prototypes used very little energy, the energy escape as the heat was unnoticeable. We observed the working of our prototypes under a thermal camera and barely notice any changes in their temperature. The energy usage of our smart textiles was also scalable, meaning that miniature structures will consume even lesser energy so does the escaping heat. We have mathematically developed and experimentally validated analytical models to establish the elongation-force relationship of the knitting and weaving configurations. We were unable to provide an analytical model for bending actuators due to the complexity when incorporating the base material specifications and properties and also the contact status between the AMF and the base.

We also introduced several programmable techniques to create multimodal motion of smart textiles. With suitable constraints, the smart textiles’ ordinary area expansion could be transformed into an s-shape, hyperbolic paraboloid, hollow structure, compression sleeve, bidirectional bending, independently bending and elongating manner. Our smart textile configurations offer versatility, programmability, and shape-shifting capabilities, which are favorable characteristics in the soft robotic field for the development of smart garments, wearable devices, and bio-inspired shape-morphing structures.

In the second approach, we embedded the AMFs into conventional passive fabrics to create desired bending motions and deformations upon pressurization. This simple, fast, and easy method facilitated shape-programmable filaments, whose segments could be mechanically programmed to become active or passive to generate the desired deformations. We also applied this concept to reconfigure passive planar fabric sheets into active and controllable 3D structures. We then demonstrated through several bio-inspired shape-morphing structures such as four-legged structures, a butterfly, and a flower. The ability to transform passive 2D sheets into active 3D structures with desired motions using AMFs opens a new possibility to manipulate objects for use in various domains. For instance, compact and deployable devices for space missions, construction, and industry. It can also be used to create bio-inspired soft robots that provide biomimetic motions for decorations and fashion design as well as wearable assistive devices for haptic display, rehabilitation, and human augmentation.
We have investigated a wide range of shape-shifting capabilities of AMF-based smart textile configurations focusing on their motions (or deformations) rather than their generated force. Although we have introduced different approaches to create different smart textiles for different applications, these works are proof-of-concepts that we believe will provide an alternative technology to the soft robotics-driven smart textiles for many applications, ranging from smart compression garments to shape-shifting robots for search and rescue, or exhibition. Despite advances, several areas need to be improved. Therefore, we suggest that future work should focus on studying comprehensively a particular smart textile configuration in terms of motion, force, and specific applications. Analytical models for bending actuators are desired for control purposes. Another area that could improve the performance of the smart textile is the use of nonlinear hysteresis modeling and adaptive control where real-time output feedback from a soft sensor will be used for closed-loop control [53, 54]. Since AMFs can be made meter-long, they can possibly be used as filaments in small-scale knitting, weaving, and embroidering machines to produce smart textiles with desired specifications and high reliability. A study to demonstrate the feasibility of machine-made smart textiles and washable smart textiles is essential to speed up the smart garment industry. For example, our artificial muscle can be used with advanced and automated sewing or embroidery machine such as the ones from Tajima Industries Ltd, Japan to form large-scale smart textile. It is desirable to develop a wearable hydraulic power source to drive smart textiles for further applications. During the experiments, we also noticed that there was a rotational motion of individual AMFs which is due to the nature of the helical coil as the outer constraint layer. However, this rotation slightly affected the elongation of the smart textile. In some configurations where the AMFs were arranged in opposite directions, the rotational effect is unlikely. For some applications where a purely linear extension is highly desired, mathematical models which take into account this effect are highly desired. We observed through demonstrations that the pressure loss was insignificant and the energy conversion was effective even with meter-long AMFs. However, we suggest a comprehensive study on the relationship between pressure loss and AMFs’ specifications to benefit precise control strategies. Finally, further research is required to explore the vast potential of incorporating functional components into smart textiles to provide additional benefits (sensing, variable stiffness) alongside the fundamental actuation.

In conclusion, this article introduces a new class of smart textiles constructed from various configurations of fluid-driven artificial muscles. The proposed smart textiles provide a high degree of versatility and programmability, enabling new possibilities in the soft robotic field,
including shape-shifting structures, biomimicking soft robots, locomotion robots, and smart garments. We anticipate this concept will inspire related improvement and development, benefiting the robotic field as a whole.

6.5. Methods

6.5.1. Material selections and specifications of smart textile prototypes

We used two types of AMFs to configure smart textiles. AMFs type S were made of silicone rubber tube OD 1.19 mm, ID 0.64 mm (Saint-Gobain, France) and stainless steel coil OD 1.49 mm, wire diameter 0.17 mm (Asahi Intecc, Japan). AMFs type M were made of latex rubber tube OD 3.18 mm, ID 1.59 mm (McMaster-Carr, USA) and stainless steel coil OD 3.18 mm, wire diameter 0.33 mm (McMaster-Carr, USA).

The knitting sheet (Fig. 6.2A) has a dimension of 54 × 52 mm constructed from three L300 mm AMFs type S using the weft knitting technique with three courses and seven wales. The bidirectional weaving sheet (Fig. 6.2B) has a dimension of 52 × 52 mm constructed from two L400 mm AMFs type S using the plain weaving technique with a pattern of 7 × 7 lines. The unidirectional weaving sheet (Fig. 6.2C) has a dimension of 57 × 27 mm constructed from one L400 mm AMF type S as warp with seven lines and acrylic yarns as weft. The circular weaving structure (Fig. 6.2D) has OD 58 mm, ID 18 mm constructed from one L500 mm AMF type M. The three bending actuators (Fig. 6.3B) have their active-section lengths of 20, 30, and 50 mm constructed from three AMFs type S with the corresponding lengths of 60, 80, and 120 mm. The four-legged structure (Fig. 6.3C) has a dimension of 60 × 50 mm constructed from one L200 mm AMF type S routing the boundary of a piece of rectangular fabric (cotton muslin fabric). The unidirectional weaving sheet (Fig. 6.6C) has a dimension of 130 × 80 mm constructed from one L1200 mm AMF type S as warp with nine lines and cotton yarns as weft. The unidirectional weaving sheet (Fig. 6.6D) has a dimension of 140 × 72 mm constructed from one L920 mm AMF type S as warp with six lines and acrylic yarns as weft.

6.5.2. Experimental setup for smart textile characterization

The testing platform consisted of a motorized linear slider (Zaber, Canada) that drove a medical syringe (BD Biosciences, Canada) to provide input volume (distilled water) and pressure to a specimen (Supplementary Fig. 6.S1). A pressure sensor (Honeywell, USA) was located right after the syringe outlet to measure input pressure. In the cases of knitting and weaving
specimens, their distal ends were connected to a linear slider and an encoder (US Digital, USA) to measure output displacement. A thin elastic string was used to prevent slack when collecting displacement data. The encoder set was replaced with a load cell (Futek, USA) when measuring the contraction force. In the cases of radial expansion and bending actuators, a digital camera was located on top of the platform to capture the specimens’ deformation.

We connected the syringe outlet to all three AMFs of the knitting sheet via a 1-to-3 fluid distributor. In the elongation tests, we drove the syringe plunger to supply hydraulic pressure to the specimen by a 0.1 Hz sinusoidal signal with an amplitude that generated a maximum pressure of 1 MPa. In the force tests, we supplied hydraulic pressure to the specimen until reaching 1 MPa and then connected the specimen’s distal end to the load cell. Subsequently, we withdrew the syringe plunger by a 0.1 Hz sinusoidal signal with an amplitude that kept the minimum hydraulic pressure at 0.1 MPa. We also applied this testing procedure for each AMF of the weaving sheet.

For radial expansion experiments, we gradually increased input pressure to the specimen until reaching a maximum pressure of 1 MPa and then reduced pressure to the initial stage at the same speed. The specimen’s deformation during the testing procedure was recorded by the camera and processed afterward to obtain the area changes. We applied a similar procedure to measure the angle changes of the bending actuators.

References


Appendix: Supplementary Information

6.S1. Experimental setup for smart textile characterization

![Experimental setup for smart textile characterization](image)

**Figure 6.S1.** Experimental setup to measure output elongation (A) and force (B) of specimens when receiving hydraulic pressure.

6.S2. Characteristics of smart textiles regarding input volume

Since all specimens were made of artificial muscle fibers (AMFs), the overall trend of experimental results (Fig. 6.S2) agrees with the AMF’s fundamental characteristic where the input volume has a proportional relationship with output elongation and a reverse proportional relationship with contraction force.

Each AMF of the weaving sheet received 0.2 mL input fluid volume to generate approximately 30% elongation (Fig. 6.S2A). The volume-elongation hysteresis chart showed a narrow gap between the pressurizing and releasing phases, indicating a good response of AMFs to hydraulic in both expansion and contraction motions. (Fig. 6.S2A). The weaving sheet could exert a contraction force of 5.6 N after releasing input volume and pressure (Fig. 6.S2B). The volume-force hysteresis chart also showed a small gap between the releasing curve and the pressurizing curve. The weaving sheet’s area expansion relies on the volume amplitudes supplied to each of the two AMFs, which are shown in the three-dimensional surface plot (Fig. 6.S2C). Experiments also revealed that the weaving sheet could generate an area expansion of 66% when its warp and weft AMFs simultaneously received 0.2 mL fluid volume.
Figure 6.2. Characteristics of smart textile configurations. (A, B) Hysteresis profiles of input volume and output elongation and force of the weaving sheet. (C) Area expansion of the weaving sheet. (D, E) Relationship between input volume and output elongation and force of the knitting sheet. (F) Area expansion of the radial expansion structure. (G) Bending angles of three different lengths of bending actuators.

Experimental results of the knitting sheet have a similar pattern to that of the weaving sheet, including the narrow hysteresis gap in the volume-elongation chart. Interestingly, the releasing curve overlapped the pressurizing curve in the volume-force chart. The knitting sheet generated 30% elongation when receiving 0.43 mL fluid volume and subsequently exerted a contraction force of 9 N (Fig. 6.2D and 6.2E).

In the case of the circular weaving sheet, there was an expansion from its initial area to 25% after receiving 0.7 mL fluid volume (Fig. 6.2F). There was a dead zone of input volume until 0.14 mL before the specimen started to expand. This dead zone is created by the initial tension of the AMF. Fig. 6.2F also showed that the releasing curves almost overlapped the pressurizing curves, denoting inconsiderable energy loss when switching the circular sheet motions.

Experimental results for three bending actuators (fabric reconfiguration) showed that their hysteresis profiles share similar patterns (Fig. 6.2G), in which they underwent a short dead zone before rising. We supplied the same volume of fluid (0.035 mL) to the three bending actuators (L20, L30, and L50 mm). However, each actuator generated different bending angles.
The L20 and L30 mm actuators reached a bending angle of 167° and 194°, respectively. The longest bending actuator (L50 mm) achieved the largest bending angle of 236°. The volume-angle hysteresis chart also showed a relatively narrow gap between the pressurizing and releasing curves of all three bending actuators.

6.3. Analytical models to establish the elongation-force relationship

**Knitting sheet**

Considering a knitting sheet made of \( n \) similar AMFs (or \( n \) courses), each has an outer diameter \( d \) and forms \( m_k \) loops (or wales) (Fig. 6.3A). Initially, the knitting sheet had a dimension of \( h_{ki} \times v_{ki} \) in horizontal and vertical directions, respectively. It was assumed that the AMFs were uniformly knitted, resulting in analogous loops. Each typical loop was identified by the width \( g_i (g_i = h_{ki}/m_k) \), length \( u_i \), and looping angle \( \phi_i \) at the initial phase. There were overlaps between two adjacent courses, whose width was approximately \( g_i/2 \). Relationships between these initial parameters are described as follows:

\[
v_{ki} = nu_i - (n - 1)\frac{g_i}{2}
\]

\[
\Rightarrow u_i = \frac{2v_{ki} + (n - 1)g_i}{2n}
\]

\[
\phi_i = \tan^{-1}\frac{g_i}{2(u_i - g_i)}
\]

When applying pressure \( P_k \) to all AMFs, the knitting sheet elongated by \( \varepsilon_{kv} \) in the wale direction and reached a new length \( v_{kp} = v_{ki}(1 + \varepsilon_{kv}) \). Subsequently, it exerted a contraction force \( F_{kv} \) when connecting to an external load and releasing input pressure. We observed that the knitting sheet was stretched in the wale direction and shrunk in the course direction simultaneously when working against an external load. \( \mu \) is the ratio between the course strain (\( \varepsilon_{kh} \)) and the wale strain (\( \varepsilon_{kv} \)), \( \mu = \varepsilon_{kh}/\varepsilon_{kv} \), and is called the Poisson’s ratio of the knitting sheet. Experiments revealed \( \mu \) was closely related to the knitting loop aspect ratio, \( \mu \approx g_i/u_i \).

At the stretching phase, the knitting sheet had a new dimension of \( h_{kp} \times v_{kp} \), where \( h_{kp} = h_{ki}(1 - \mu\varepsilon_{kv}) \) and \( v_{kp} = v_{ki}(1 + \varepsilon_{kv}) \). Also, each knitting loop had a new width \( g_p (g_p = h_{kp}/m_k) \), length \( u_p \), and looping angle \( \phi_p \).
Chapter 6. Smart textiles made from hydraulic artificial muscles

\[ u_p = \frac{2v_{kp} + (n - 1)g_p}{2n} \]  

(S3)

\[ \varphi_p = \tan^{-1} \left( \frac{g_p}{2(u_p - g_p)} \right) \]  

(S4)

Each loop consisted of two legs (Fig. 6.S3A), whose length increased from \( l_{ki} \) at the initial phase to \( l_{kp} \) for the stretching phase, where \( l_{ki} = (u_i - g_i) / \cos \varphi_i \) and \( l_{kp} = (u_p - g_p) / \cos \varphi_p \). \( x_{kv} \) is the displacement of each leg, namely \( x_{kv} = l_{kp} - l_{ki} \).

The maximum contraction force \( F_{out} \) of an AMF (when input pressure \( P = 0 \)) was given in reference [49], and is reintroduced as follows:

\[ F_{out} = \alpha E A_0 \left( 1 - \frac{1}{1 + x/l_i} \right) + kx \]  

(S5)

where \( \alpha, E, A_0 \) represents the stretch ratio, Young’s modulus, and cross-sectional area of the silicone tube, respectively; \( k \) is the stiffness coefficient of the helical coil; \( x \) and \( l_i \) are the displacement and initial length of the AMF, respectively.

**Figure 6.S3.** Analytical model illustration to establish the elongation-force relationship. (A) Knitting sheet. (B) Weaving sheet.
The last constituent $kx$ in Eq. (S5) describes the helical coil elastic force $F_c$ which is assumed to be linear-elastic or Hookean. However, the accumulative initial tension of helical coils in knitting and weaving configurations must be considered. Therefore, we propose a comprehensive equation to describe the coil force, as shown in Eq. (S6). The coil force equation consists of a short linear segment at the first $2\%$ strain followed by a modified Hookean segment that accounts for the initial tension.

$$F_c(x) = \begin{cases} qx, & \varepsilon \leq 2\% \\ kx + F_0, & \varepsilon > 2\% \end{cases}$$

(S6)

Continuity condition: $q = k + F_0/x$, $\varepsilon = 2\%$

where $\varepsilon$ and $F_0$ are the strain and initial tension of the helical coil, respectively.

Since the knitting sheet works against an external load in the wale direction, its contraction force is equivalent to that of a single course. There are $2m_k$ legs for each course. The contraction force of each leg follows Eq. (S5), thus the knitting sheet contraction force can be expressed as:

$$F_{kV} = 2m_k \left[ \alpha E A_0 \left( 1 - \frac{1}{1 + \frac{1}{x_{kV}/l_{kt}}} \right) + \begin{cases} qx_{kV}, & \varepsilon_{kV} \leq 2\% \\ kx_{kV} + F_0, & \varepsilon_{kV} > 2\% \end{cases} \cos \varphi_p \right]$$

(S7)

Continuity condition: $q = k + F_0/x_{kV}$, $\varepsilon_{kV} = 2\%$

Weaving sheet

Consider a weaving sheet made of two AMFs (H and V, each has an outer diameter $d$) that follows a $m_h \times m_v$ lines criss-cross pattern (Fig. 6.S3B). The weaving sheet dimension is $h_{wi} \times v_{wi}$ at the initial stage. It is assumed that the distance between two adjacent lines is identical for each AMF. Thus, the distance between two lines of the AMF H is $s_{vi} = v_{wi}/(m_h - 1)$ and the distance between two lines of the AMF V is $s_{hi} = h_{wi}/(m_v - 1)$. Since two AMFs are similar, we solely present here the equation derivation for the AMF H. In plain weave, when a line crosses over or under another line, it forms an angle with the transverse orthogonal plane. We refer to this angle as an interlocking angle, $\theta_{hi}$ for lines of the AMF H.

$$\theta_{hi} = \tan^{-1} \frac{d}{s_{hi}} = \tan^{-1} \frac{d(m_v - 1)}{h_{wi}}$$

(S8)
The AMF H has \( m_h \) lines with the same length \( l_{hi} = h_{wi}/\cos\theta_{hi} \). When applying pressure \( P_h \) to the AMF H, the weaving sheet elongates \( \varepsilon_{wh} \) horizontally and reaches a new length \( h_{wp} = h_{wi}(1 + \varepsilon_{wh}) \) and a new interlocking angle \( \theta_{hp} \).

\[
\theta_{hp} = \tan^{-1}\left(\frac{d(m_v - 1)}{h_{wp}}\right) = \tan^{-1}\left(\frac{d(m_v - 1)}{h_{wi}(1 + \varepsilon_{wh})}\right)
\] (S9)

The new length of weaving lines is \( l_{hp} = h_{wp}/\cos\theta_{hp} \). Let \( x_{wh} \) be the displacement of a line, given as \( x_{wh} = l_{hp} - l_{hi} \). Adapting to Eq. (S7), the weaving sheet contraction force in direction H can be expressed as:

\[
F_{wh} = m_h \left[ \alpha E A_0 \left(1 - \frac{1}{1 + x_{wh}/l_{hi}}\right) + \left\{ qx_{wh}, \quad \varepsilon_{wh} \leq 2\% \right\} + \left\{ kx_{wh} + F_0, \quad \varepsilon_{wh} > 2\% \right\} \right] \cos\theta_{hp}
\] (S10)

Continuity condition: \( q = k + F_0/x_{wh}, \quad \varepsilon_{wh} = 2\% \)

6.54. Shape-programmable filaments

We have developed a simple technique to create a bending actuator – by attaching an AMF to a strip of non-stretchable fabric. This section extends the bending actuator concept to shape-programmable filaments, in which we can strategically allocate multiple active and passive segments in a single AMF to produce the desired shape.

We first attached an AMF to a strip of non-stretchable fabric by a layer of double-sided tape, creating a single active filament. Later, we deliberately added another layer of tape at locations that we intended to be passive (Fig. 6.54A). Note that multiple strips of fabric can be attached to one AMF. Also, the length and direction of these strips vitally affect bending curvature and direction of the active segments. Fig. 6.54B and 6.54C demonstrate four active filaments have been programmed to transform their shapes from straight lines into letters (UNSW) upon pressurization. This simple technique enables the shape-shifting capability of AMFs to turn 1D lines into 2D shapes and possibly 3D structures.
Figure 6.54. Shape-programmable filaments. (A) Attaching an AMF to a strip of non-stretchable fabric to create a bending actuator. (B, C) Programming four bending actuators to form four letters (UNSW) upon pressurization.
Chapter 7

Modeling and control of woven hydraulic artificial muscles

This chapter extends the study of smart textiles in Chapter 6 in terms of modeling, control and application. Specifically, the fabrication process of a woven hydraulic artificial muscle (a smart woven textile) for a wearable application is presented. A simple nonlinear hysteresis model based on the modified Bouc-Wen model is developed and implemented into a feedforward control scheme to capture the output elongation of the developed smart woven textile. Furthermore, an adaptive controller is proposed to further improve output tracking performance. Both controllers exhibited remarkable performance in reducing tracking errors compared to the uncompensated system. This chapter also presents the development of a portable smart compression device to deliver active compression therapy for the treatment of venous disease or compromised peripheral blood flow. The proposed compression device is compact, portable, easy to put on/take off, and can be controlled wirelessly using a user-friendly smartphone application. Experiments revealed that the compression device could provide a comparable compression force and speed with low energy consumption. The developed smart woven textile can incorporate a soft contact force sensor and other electronic components for additional functionality.

This work has been published:

7.1. Abstract

Soft robotics, soft artificial muscles and smart textiles are on the rise to compete with their rigid counterparts in the fields of wearable technologies and medical applications owing largely to their improved compliance characteristics. This paper introduces a smart textile created by weaving a hydraulic artificial muscle for a wearable application. To capture the inherent nonlinear hysteresis of the developed smart woven textile, we propose a simple hysteresis model and implement a feedforward compensation scheme and an adaptive controller. Both control strategies significantly reduced the output elongation tracking errors, demonstrating up to 90% improvement in the case of adaptive control compared to the uncompensated system. We also present the development of a portable smart compression device to overcome the shortcomings of current compression devices to provide treatment for people with venous disease or compromised peripheral blood flow. While current compression devices are passive, difficult to put on/take off, slow to compress or noisy and bulky, our smart compression device is easy to wear, compact and portable, with low energy consumption and can be easily controlled with a smartphone application via Bluetooth connection. The smart compression device operated at 18% strain and took 5 s to complete each compression cycle. We also demonstrate the proof-of-concept of incorporating a soft contact force sensor and other electronic components into the smart textile structure for extra functionality. This work contributes to the advancement of soft robotic technologies, wearable electronics and smart wearables for medical applications.

7.2. Introduction

Soft robotics is a prolific area of research that has recently gained substantial traction in the robotic community. In particular, soft actuators or soft artificial muscles have emerged as potential candidates to replace rigid actuators in a wide range of applications [1]. Compared to their rigid counterparts, soft actuators are compliant, resilient to disturbance and can provide human-friendly interaction, making them well-suited for wearable applications and use in medical devices [2, 3]. Soft actuators are generally categorized based on their excitation sources such as electric, thermal, chemical, magnetic or pressure [4]. Electrically-driven soft actuators are usually made of electroactive polymers (EAPs) whose strain varies under electrical excitation. A well-known representative of electrically-driven actuators is dielectric elastomer actuators (DEAs), which can produce moderate strain and fast actuation at the trade-
off of an extremely high operating voltage (several kilovolts) [5, 6]. Thermally-driven soft actuators such as shape memory polymers (SMPs) and twisting artificial muscle filaments show impressive performance in terms of high strain and high stress but have the intrinsic drawbacks of low energy efficiency and slow response [7, 8]. Chemically-driven soft actuators rely on the principle of chemomechanics, where the energy generated from chemical reactions is transformed into mechanical energy. For example, Nafion membranes restored their shape through solvent absorption [9]. Most chemically-driven soft actuators experience slow response times [4]. Magnetically-driven soft actuators are made of elastomers mixed with magnetic particles so that their motion can be controlled by an external magnetic field. They are fast and programmable but have limited actuation force [10, 11].

Pressure-driven soft actuators receive pressure power (pneumatic or hydraulic) to generate shape changes and exert actuation force. They have simple structures but demonstrate excellent performance including high strain, high stress and high energy efficiency [12, 13]. The Suzumori group has developed the famous pneumatic artificial muscle called the McKibben actuator, which is made of an inner elastic tube and an outer mesh [14-16]. The McKibben actuator expands radially and contracts axially when receiving positive pneumatic pressure. With an air pressure of 0.5 MPa, it contracted 34% of its initial length and exerted a contraction force of 15.77 N [17]. Hawkes et al. introduced an inverse pneumatic artificial muscle made from an elastic tube with reinforced filaments that produced over 300% strain when pressurizing [18]. In another approach, Li et al. developed a vacuum artificial muscle made from an origami structure sealed inside a bag that exerted 90% strain [19]. The Do group developed a hydraulic filament artificial muscle from an inner silicone tube and an outer coil for multiple robotic and medical applications [12, 20, 21]. This technology had a high length-per-diameter ratio and could provide 246.8% strain upon hydraulic pressurization.

After extensively exploring possibilities for creating soft artificial muscles, many researchers have shifted their interests to enhancing the performance and usability of artificial muscles through structural configurations such as twisting, braiding, knitting and weaving [15, 22, 23]. As an inevitable result, smart textiles have arisen and have quickly become a fertile research topic worldwide. Medical demand is also a strong driving force to accelerate the development of smart textiles, especially in the field of wearable assistive devices. For example, compression devices that provide compression force to the human body to prevent blood from pooling and promote circulation are highly sought after for the treatment of postural orthostatic tachycardia syndrome (POTS), lymphedema, and for the prevention of venous
thromboembolism (VTE) [24, 25]. The very first attempt to treat these conditions used passive garments or graduated compression stockings, which caused discomfort and skin irritation to the patients [26]. In contrast, active garments with dynamic compression are more favorable. Payne et al. created a pneumatically inflatable garment with adjustable compression force for mechanotherapy [27]. Yang et al. developed a fabric-based active compression sleeve powered by a tendon mechanism that could change and maintain compression force [28]. In another approach, Pettys-Baker et al. used shape memory alloy (SMA) coil actuators to create an active compression garment to treat orthostatic intolerance [29]. Current active compression devices are noisy with bulky pneumatic actuation sources or have limited compression rates. Therefore, there is a huge demand for a smart compression device with adjustable and dynamic compression, easy to put on and take off as well as being low noise and portable.

Smart textiles benefit from the flexibility and conformability of conventional passive textiles as well as having the controllability of soft artificial muscles. Smart textiles can be created by either integrating soft artificial muscles into fabrics that act as a constrained layer or interlacing soft artificial muscle filaments to form smart sheets [30-32]. In the former approach, the fabric layer restrains certain deformation of the enclosed elastic components, resulting in specific motion (elongating, twisting, bending) of the smart textiles upon excitation. Zhu et al. proposed fluidic fabric muscle sheets made by inserting a silicone tube inside prefabricated fabric channels to induce various shape-shifting motions when hydraulically pressurized [33]. Similarly, Booth et al. developed OmniSkins, a smart sheet that could wrap around passive objects and turn them into multifunctional active robots [34]. In the latter approach, long artificial muscle filaments served as active yarns to construct smart textiles using traditional knitting or weaving techniques. Maziz et al. knitted and weaved electrically responsive polymer filaments to form electroactive textiles for wearable devices [23]. Hiramitsu et al. used thin McKibben muscles to create active sheets that could exert high contraction force and moderate elongation [35]. In previous work, we introduced smart textiles made from fluid-driven artificial muscle fibers using knitting and weaving techniques [36]. Our smart textiles could be mechanically programmed to induce multimodal motion and shape-shifting.

The compliance properties of soft materials compromise the control aspect of soft actuators. Compared with their rigid counterparts, the inherent nonlinear elastic hysteresis of soft actuators significantly increases the complexity of control strategies [37]. The situation is made worse in the case of smart textiles due to the accumulation of hysteresis of the soft actuators. Hysteresis is a memory-dependent and highly nonlinear behavior [38]. The area within the
hysteresis loops corresponds to the amount of energy lost due to internal friction. These nonlinearities greatly complicate the position control of soft actuators since the output cannot be known without knowing the current state of the system. This behavior highlights the need for nonlinear hysteresis models and advanced control techniques such as feedforward and adaptive control.

There are broadly two types of hysteresis models. The first are operator-based models that employ hysteresis factors with slope-specific weights of the loading and unloading phases. Some well-studied models in this category include the Preisach model and the Prandtl-Ishlinskii model [39, 40]. The second type of hysteresis model is a differential-based model that describes the behavior of a hysteretic system using nonlinear differential equations. Perhaps the most popular model used to describe hysteresis-like behavior is the Bouc-Wen model, known for its versatility and computational simplicity [41, 42]. Do et al. proposed a modified Bouc-Wen model to capture the asymmetric hysteresis loop of the tendon-sheath mechanism [43, 44]. In the field of soft robotics, Thai et al. developed a hysteresis model based on the Bouc-Wen model to describe the asymmetric hysteresis attribution of a hydraulic artificial muscle and then demonstrated it in a miniature robotic catheter [45].

![Smart woven sheet for wearable applications](image)

**Figure 7.1.** Smart woven sheet for wearable applications.

This paper addresses some of the shortcomings mentioned above. In detail, we introduce a smart textile made from a single hydraulic artificial muscle and conventional yarn using traditional weaving techniques. We then employ the Bouc-Wen model to describe its hysteresis.
behavior. Subsequently, the proposed model is inversed and implemented in a feedforward controller to track the output elongation of the smart textile. We also investigate and apply an adaptive controller to enhance tracking performance. Furthermore, we demonstrate a portable smart compression device that allows users to control an active compression sleeve through a smartphone with an actuation unit on their waist (Fig. 7.1). The smart compression device is also capable of carrying contact sensors and other electronic components for additional functionality.

7.3. Woven hydraulic artificial muscles

7.3.1. Design, fabrication and working principle

The smart woven sheet or the woven hydraulic artificial muscle (WHAM) is made by the centuries-old plain weave technique where a long hydraulic artificial muscle (HAM) and conventional yarn are interlaced at right angles to form a textile (Fig. 7.1). The HAM consists of an inner silicone tube to receive fluid power and an outer coil spring to constrain the silicone tube’s radial expansion. One end of the HAM is blocked while the other end is connected to a fluid source (e.g. syringe) via a fluid transmission tube. To minimize the number of required fluid sources, a single long HAM is arranged in a meandering course to serve as multiple lines of the warp while conventional yarn plays the role of the weft to fill and stabilize the WHAM.

Due to the radial constraint of the coil spring, HAM will extend longitudinally and accumulate elastic energy when receiving positive hydraulic pressure from the fluid source. Subsequently, the HAM will discharge this elastic energy and return to its initial length when the pressure is released. Therefore, we can control the HAM’s elongation by regulating the applied hydraulic pressure. The HAM’s elongation directly generates the WHAM’s elongation in the warp-wise direction. Each line of the warp equally contributes to the WHAM’s elongation with the assumption of the HAM’s uniformity.

We fabricated a WHAM prototype called WHAM1 with a dimension of 137 mm in length and 40 mm in width (L137 × W40 mm) from a HAM with an outer diameter of 1.5 mm and a length of 1200 mm (OD1.5 × L1200 mm) and 4-ply cotton yarn (Fig. 7.1). The HAM was arranged in eight equal lines 6 mm apart while the cotton yarn was woven with a relatively high density. Two ends of the WHAM1 were clamped by plastic 3D-printed parts to facilitate later usage. Detailed specifications of the prototype can be found in Table 7.1.
7.3.2. Experimental characterization

We built an experimental platform to measure the WHAM1’s characteristics (Fig. 7.2A). The platform consisted of a motorized linear slider (Faulhaber, Germany) that drove a syringe plunger to supply hydraulic power (e.g. water) to the WHAM1. The WHAM1’s proximal end was fixed while its distal end was connected to an encoder (US Digital, USA) via a linear slider to measure displacement. We provided input signals to the motorized linear slider and received output displacement from the encoder using a central control unit comprising a computer, MATLAB Simulink (MathWorks, USA) and a data acquisition board (Quanser, Canada).

**Table 7.1. Specifications of prototypes**

<table>
<thead>
<tr>
<th></th>
<th>Woven hydraulic artificial muscle 1 (WHAM1)</th>
<th>Fabric-based hydraulic artificial muscle (FHAM)</th>
<th>Woven hydraulic artificial muscle 2 (WHAM2)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Coil spring</strong></td>
<td>Stainless steel (Asahi Intecc, Japan)</td>
<td>Carbon steel (AliExpress, China)</td>
<td>Carbon steel (McMaster-Carr, USA)</td>
</tr>
<tr>
<td></td>
<td>OD: 1.5 mm</td>
<td>OD: 2.5 mm</td>
<td>OD: 3.2 mm</td>
</tr>
<tr>
<td></td>
<td>ID: 1.2 mm</td>
<td>ID: 1.9 mm</td>
<td>ID: 2.5 mm</td>
</tr>
<tr>
<td></td>
<td>L: 1200 mm</td>
<td>L: 1060 mm</td>
<td>L: 914 mm</td>
</tr>
<tr>
<td></td>
<td>k: 8.15 N/m</td>
<td>k: 21.4 N/m</td>
<td>k: 68 N/m</td>
</tr>
<tr>
<td><strong>Silicone tube</strong></td>
<td>Silicone rubber (Saint-Gobain, France)</td>
<td>Silicone rubber (Saint-Gobain, France)</td>
<td>Latex rubber (McMaster-Carr, USA)</td>
</tr>
<tr>
<td></td>
<td>OD: 1.2 mm</td>
<td>OD: 2.2 mm</td>
<td>OD: 3.2 mm</td>
</tr>
<tr>
<td></td>
<td>ID: 0.6 mm</td>
<td>ID: 1 mm</td>
<td>ID: 1.6 mm</td>
</tr>
<tr>
<td></td>
<td>E (100%): 1.648 MPa</td>
<td>E (100%): 1.586 MPa</td>
<td>E (100%): 1.856 MPa</td>
</tr>
<tr>
<td><strong>Filler</strong></td>
<td>4-ply cotton yarn (Kmart, Australia)</td>
<td>Self-adhesive compression bandage (3M, Australia)</td>
<td>8-ply acrylic yarn (Kmart, Australia)</td>
</tr>
</tbody>
</table>

k: spring constant; E (100%): Young’s modulus at 100% strain

We supplied hydraulic power (degassed water) to the WHAM1 using a 0.05 Hz sinusoidal signal with peak-to-peak amplitudes of either 0.26 mL or 0.52 mL. Respectively, WHAM1 reached its peak elongations at 16.7% and 34.3% (Fig. 7.2B). The elongation-volume relationship is proportional and relatively linear with a slope of approximately 65 %/mL.
However, hysteresis exists between the releasing phase and the pressurizing phase, where a gap of around 0.02 mL separates the two phases (Fig. 7.2B). The hysteresis gap reflects energy loss when the WHAM1 changes its motion from elongation to contraction.

**Figure 7.2.** Experimental characterization of WHAM1. (A) Experimental setup. (B) Relationship between output elongation and input volume with two different amplitudes (0.26 mL and 0.52 mL) at 0.05 Hz. (C) Relationship between output elongation and input volume with small amplitude (0.035 mL) at various frequencies.

We also examined the elongation-volume relationship of the WHAM1 at various frequencies ranging from 0.1 Hz to 0.6 Hz with a smaller amplitude of 0.035 mL. Experimental results are shown in Fig. 7.2C, where hysteresis profiles share similar patterns to those in the case of larger amplitudes. However, a smaller amplitude (0.035 mL) is associated with a smaller hysteresis gap (<0.01 mL). Regarding motion frequency, higher frequencies cause wider hysteresis gaps (Fig. 7.2C). Due to the long length and small diameter of the HAM, the WHAM1 exhibited poorer performance in response to higher frequencies than lower ones. It is worth noting that the WHAM1 experienced a backlash-like hysteresis, similar to that seen in the tendon-sheath system.
mechanism [46]. This phenomenon occurs due to the accumulation of nonlinear characteristics of multiple warp lines of the WHAM1, meaning that the input volume is required to reach a certain threshold before driving changes in the output elongation.

7.4. Modeling and control

7.4.1. Nonlinear hysteresis model

We employed the Bouc-Wen model to capture the WHAM1’s hysteresis. The Bouc-Wen model, which is known for its versatility and computational simplicity, is one of the most popular models used to describe hysteresis-like behavior. It is a differential-based model which describes the behavior of a hysteretic system using a nonlinear differential equation. The output $x_{out}$ is a linear function of the input $x_{in}$ and an internal state $z$, as shown in Eq. (1). The internal state $z$ describes the nonlinear behavior, Eq. (2). Coefficients $\alpha_z$, $\alpha_x$, $A$, $\beta$, $\gamma$, and $n$ dictates the size and shape of the hysteresis loop.

$$
\begin{align*}
    x_{out} &= \alpha_x x_{in} + \alpha_z z \\
    \dot{z} &= A\dot{x}_{in} - \beta|\dot{x}_{in}|z^{n-1}z - \gamma \dot{x}_{in}|z|^n
\end{align*}$$

We chose $n = 1$. This yielded a much simpler Bouc-Wen model structure (Eq. (3), (4)) which greatly simplifies the process of inverting the model and reduces the time taken to identify the model parameters while not compromising its accuracy [47].

$$
\begin{align*}
    x_{out} &= \alpha_x x_{in} + \alpha_z z \\
    \dot{z} &= A\dot{x}_{in} - \beta|\dot{x}_{in}|z - \gamma \dot{x}_{in}|z|
\end{align*}$$

Next, we need to find an optimal set of parameters that match the hysteresis model to the actual system. We identified model parameters using particle swarm optimization (PSO) – an algorithm that exploits the computational intelligence of nature [48]. PSO involves placing a number of particles within the search space of an optimization problem and evaluating the fitness at their given locations [49]. Each particle will move based on knowledge of its current position and location of highest fitness. Eventually, the swarm will converge upon an optimal fitness function, the location which corresponds to an optimal set of model parameters.

We utilized experimental data in Section 7.3 to feed the PSO algorithm (MATLAB script). A pool of 400 particles was created and allowed to move around a 5-dimensional solution space until an optimal solution was reached. The particle position cost was given by the sum of the squared error signal between the output of the Bouc-Wen model (Eq. (3), (4)) and the actual
output (WHAM1’s elongation). The PSO algorithm reached its minimum cost around the 30\textsuperscript{th} iteration with the final set of model parameters presented in Table 7.2.

Figure 7.3 compares the output of the modified Bouc-Wen model with the behavior of the actual system at an aperiodic input signal of 0.05 Hz and $0.05\sqrt{3}$ Hz. It shows the overlap between the identified model and experimental data for the majority of the observed timeframe besides some slight errors as the input changes direction. Regarding model accuracy, the identified model has a root mean square error (RMSE) of 0.0661\% and a peak-to-peak error (P2PE) of 0.2925\%.

<table>
<thead>
<tr>
<th>Table 7.2. Model parameters</th>
</tr>
</thead>
<tbody>
<tr>
<td>Identified model</td>
</tr>
<tr>
<td>----------------------------</td>
</tr>
<tr>
<td>1.541</td>
</tr>
<tr>
<td>Adaptive controller</td>
</tr>
<tr>
<td>3.5</td>
</tr>
</tbody>
</table>

\textbf{Figure 7.3.} Comparison between identified model and experimental data at an aperiodic input signal of 0.05 Hz and $0.0\sqrt{3}$ Hz. (A) Time-series output elongation. (B) Volume-elongation hysteresis.
7.4.2. Feedforward compensation control scheme

When feedback is unavailable, a feedforward controller can be used as a simple and rudimentary method to control the output of nonlinear systems. The feedforward compensation control scheme is therefore a great fit for our smart woven sheet (embodiment WHAM1) as it is designed to use as a simple smart compression sleeve that can provide pulse compression without feedback. Typically, feedforward controllers are based upon the principle of inverse compensation, which simply inverts the identified model of the system and places it in a cascade with the real system [50]. These controllers have a simple structure and are easy to implement.

To normalize the slope of the hysteresis profile as well as cancel the nonlinear behavior of the system, the control input $x_{ff}$ is chosen as follows:

$$x_{ff} = \frac{1}{\alpha_x} (x_{des} - \alpha_z \ddot{z})$$  \hspace{1cm} (5)

The proposed controller utilizes an internal state estimator $\hat{z}$ which approximates the value of a nonlinear term $z$ using the control input $x_{ff}$. The value of the estimated internal state is defined by the nonlinear ordinary differential equation.

$$\dot{\hat{z}} = A \dot{x}_{ff} - \beta |\dot{x}_{ff}| \ddot{z} - \gamma x_{ff} |\ddot{z}|$$ \hspace{1cm} (6)

Substituting Eq. (5) into the Bouc-Wen model described in Eq. (3), we get:

$$x_{out} = \alpha_x \left( \frac{1}{\alpha_x} (x_{des} - \alpha_z \ddot{z}) \right) + \alpha_z z = x_{des}$$ \hspace{1cm} (7)

This assumes all identified model parameters are accurate as well as the approximation of the nonlinear term $z$ i.e $z - \hat{z} \approx 0$. Here, $x_{des}$ is the desired displacement signal and $x_{out}$ is the output displacement of the system. The full feedforward control structure is shown in Fig. 7.4, where $x_{des}$ is calculated from the desired elongation $\varepsilon_{des}$ and initial length $l_0$ of the smart woven sheet ($x_{des} = \varepsilon_{des}l_0$). Similarly, $x_{out}$ is converted to the output elongation $\varepsilon_{out}$ ($\varepsilon_{out} = x_{out}/l_0$).

We implemented the feedforward controller structure (Fig. 7.4) in MATLAB Simulink and placed it in a cascade with the actual system (experimental setup in Fig. 7.2A). Model parameters are given in Table 7.2. Various input signals in terms of amplitude and frequency were applied to the testing platform to examine the performance of the proposed feedforward
controller. The results are presented in Fig. 7.5. For visualization purposes, hysteresis profiles of the uncompensated system are shown with normalized peak-to-peak gains.

In the first scenario, we generated and fed the controller with a periodic sinusoidal signal of the desired elongation depicted in Eq. (8). The signal has an amplitude of \( \varepsilon_A = 5.5\% \) (peak-to-peak amplitude 11\%), frequency \( f = 0.05 \) Hz, and initial phase of \( \varphi = -\pi/2 \) rad. In the second run, the controller was supplied with an aperiodic signal comprising two sinusoidal waveforms (Eq. (8)) with the corresponding amplitudes of 3.66\% and 1.83\% and frequencies of 0.05 Hz and \( 0.05\sqrt{3} \) Hz.

\[
\varepsilon_{\text{des}}(t) = \varepsilon_A \sin(2\pi ft + \varphi) + \varepsilon_A
\]  

Fig. 7.4. Structure of the feedforward controller with a detailed schematic of the \( \dot{z} \) estimator. The controller receives the desired elongation \( \varepsilon_{\text{des}} \) as input to generate control signal \( x_{\text{ff}} \) for the mechanical system. This structure is the schematic deployment of Eqs. (5) and (6).

In both periodic and aperiodic scenarios, the output elongation of experimental data closely matches those of the desired elongation with some slight errors where the input changes direction (Fig. 7.5A, C). These can be attributed to model uncertainty and the limited speed of the motor. The feedforward hysteresis profiles in both cases have smaller gaps and are more linear than those of the uncompensated system (Fig. 7.5B, D).
Figure 7.5. Performance of the feedforward controller. (A, B) Output elongation and hysteresis profile at periodic input signal of 0.05 Hz. (C, D) Output elongation and hysteresis profile at aperiodic input signal of 0.05 Hz and $0.05\sqrt{3}$ Hz. (E, F) Output elongation and hysteresis profile with small amplitude at various periodic frequencies.
Furthermore, we have provided the controller with periodic sinusoidal signals of small amplitude (0.73%) at different frequencies of 0.1, 0.2, 0.4 and 0.6 Hz. Results revealed that the tracking error increases with the input frequency (Fig. 7.5E). This can be attributed to the changing hysteresis profile of the system as the hysteresis model was developed to accommodate the low-frequency system configuration. Increasing the input frequency also widens the hysteresis loops and alleviates the nonlinear cancelling effect of the feedforward controller (Fig. 7.5F). However, the feedforward hysteresis profiles at all frequencies have a noticeably narrow gap compared with those of the normalized gain uncompensated system.

7.4.3. Nonlinear adaptive control scheme

To improve tracking errors, closed-loop feedback control schemes such as adaptive control have been successfully used in the context of hysteretic systems. Adaptive controllers differ from other controllers by their time-varying control parameters, which change according to the system feedback. The equations that govern this change are called adaption laws. This form of nonlinear control provides many improvements to the feedforward controller such as dynamically cancelling the system nonlinearities without prior information about the system (eliminating the need for system modeling) and rejecting disturbances \[51\]. Self-tuning adaptive controllers also take a finite amount of time to estimate control parameters, during which time reference tracking errors will be fairly prevalent.

We have developed a self-tuning adaptive controller for our smart woven sheet (embodiment WHAM1) based on the previous works of Do et al. \[51\] and Dinh et al. \[52\]. We replace the hysteretic nonlinear term \(\alpha z\) in Eq. (3) with a single disturbance term \(D\) to describe all nonlinearities and model uncertainty, as shown in Eq (9) below:

\[
\begin{align*}
    x_{out} &= \alpha x_{in} + D \\
    \Rightarrow x_{in} &= \theta(x_{out} - D), \quad \forall |D| \leq D_m
\end{align*}
\]

where \(\theta = 1/\alpha\) is the inverse slope of the hysteresis profile and \(D_m\) is the upper bound of the disturbance.

We define several intermediate variables in Eqs. (11-13) including a raw tracking error \(e\) between the output displacement \(x_{out}\) and the desired displacement \(x_{des}\), a filtered tracking error \(s\), a modified desired displacement \(x_{mod}\) and a positive constant \(\lambda\) whose value depends heavily on the sampling rate of the control system.
\[
\begin{align*}
\dot{e} &= x_{\text{out}} - x_{\text{des}} \\
\dot{s} &= e + \lambda \dot{e} \\
\dot{x}_{\text{mod}} &= x_{\text{des}} - \lambda \dot{e}
\end{align*}
\]

From these variables, we can form the control law \( x_{\text{ad}} \) as presented in Eq. (14) with its self-tuning functions as described in Eqs. (15, 16).

\[
\begin{align*}
\dot{x}_{\text{ad}} &= \hat{\theta} \left( x_{\text{mod}} - D_{m} \tanh \left( \frac{s}{\psi} \right) \right) - \kappa s \\
\dot{\theta} &= -\delta_{1} \overline{x}_{\text{mod}} s - \sigma_{1} \dot{\theta} \\
\dot{D}_{m} &= \delta_{2} \tanh \left( \frac{s}{\psi} \right) s - \sigma_{2} D_{m}
\end{align*}
\]

where \( \hat{\theta} \) and \( D_{m} \) are the estimates of \( \theta \) and \( D_{m} \), respectively; \( \kappa \) is a positive feedback constant; and \( \psi \) is the sensitivity constant of the switching term \( \tanh(s/\psi) \). The hyperbolic tangent function is used instead of the signum function to reduce chattering within the control signal.

The self-tuning functions’ operation is based on simple sensitivity methods that are used to steer the estimated parameter towards its real value based on the filtered tracking error term, \( s \). We also define positive adaption gains \( \delta_{1}, \delta_{2} \) and positive constants \( \sigma_{1}, \sigma_{2} \) which control the speed of convergence and ensure the parameter adaption is uniformly ultimately bounded \([52]\).

The variable \( \overline{x}_{\text{mod}} \) is given in Eq. (17). The complete structure of the adaptive controller is presented in Fig. 7.6.

\[
\overline{x}_{\text{mod}} = x_{\text{mod}} - D_{m} \tanh \left( \frac{s}{\psi} \right)
\]

We implemented the adaptive controller structure (Fig. 7.6) in MATLAB Simulink and placed it in a cascade with the actual system (experimental setup in Fig. 7.2A). Tuning the parameters was fairly straightforward. Firstly, \( \lambda \) was chosen based on the sampling rate of the control system. Secondly, \( \kappa \) was chosen as a relatively low value to maintain a balance between speed and stability. From here, the remaining parameters were adjusted to achieve the desired adaptation time and disturbance response time. The selection of these parameters will vary depending on the intended use of the system, e.g. a rapidly changing configuration will require fast parameter adaptations to minimize errors and vice versa. The tuned parameters are shown in Table 7.2.

To evaluate the adaptive controller performance, we generated and fed it various input signals similar to those we provided to the feedforward controller. In the low-frequency scenarios of periodic (0.05 Hz) and aperiodic (0.05 Hz and 0.05\( \sqrt{3} \) Hz) input signals, the adaptive controller
takes roughly 4 s to approximate $\hat{\theta}$ and $\hat{D_m}$ during which time the error is relatively minimal. Beyond that point, the error is almost zero for the entire experimental timeframe (Fig. 7.7A, C). As a result, the adaptive controller hysteresis loops have a remarkably narrow gap and high linearity compared with the uncompensated system (Fig. 7.7B, D).

**Figure 7.6.** Structure of the adaptive controller with detailed schematics of the $\hat{\theta}$ and $\hat{D_m}$ estimators. The controller receives the desired elongation $\varepsilon_{\text{des}}$ as input to generate control signal $x_{\text{ad}}$ for the mechanical system. This structure is the schematic deployment of Eqs. (11)-(17).

In the case of small amplitude and high-frequency input signals, the performance of the adaptive controller gradually decreases with frequency, especially when the input signal changes direction (Fig. 7.7E, F). This was anticipated due to the speed limit of the motorized linear slider as well as the sampling frequency of the data acquisition board. Overall, the adaptive controller has provided much better hysteresis loops in terms of linearity and hysteresis gap than the normalized gain uncompensated system.

Since the adaptive controller uses output feedback, it is capable of detecting changes to the system configuration in real-time and adjusting control parameters to compensate. We introduced sudden disturbances to the running system by shifting the smart woven sheet perpendicular to the elongation direction and found that the adaptive controller could effectively reject disturbances with a magnitude up to 1.2% (at 11% peak-to-peak amplitude input signal) and forced the error back to zero within 1 s. However, fast-changing disturbances beyond this amplitude can cause the adaptive controller to enter an unstable state, whereby the system oscillates rapidly at an increasing amplitude.
Figure 7.7. Performance of the adaptive controller. (A, B) Output elongation and hysteresis profile at periodic input signal of 0.05 Hz. (C, D) Output elongation and hysteresis profile at aperiodic input signal of 0.05 Hz and $0.05\sqrt{3}$ Hz. (E, F) Output elongation and hysteresis profile with small amplitude at various periodic frequencies.
7.4.4. Accuracy comparison

Figure 7.8 shows the root mean square error (RMSE) between the experimental data and the desired input signals of the feedforward controller and the adaptive controller together with the normalized gain uncompensated system for reference. It is clear that the tracking error increases with amplitude and frequency for all control schemes. Both the feedforward controller and adaptive controller show significant improvement over the uncompensated system for all tested frequencies and amplitudes. Compared to the normalized gain uncompensated system, the feedforward controller and adaptive controller reduce RMSE by up to 60% and 90% respectively. It is obvious that the adaptive controller consistently outperforms the feedforward controller at all amplitudes and frequencies. This indicates a distinct advantage of a feedback-based control scheme over a feedforward-based scheme, that is, the feedback-based algorithm can account for model uncertainty.

![Accuracy comparison](image)

**Figure 7.8.** Accuracy comparison between different control schemes at various amplitudes and frequencies.

7.5. Portable compression device

From our perspective, an ideal compression device should be easy to use, lightweight, compact and portable, enabling users to enjoy a quick session of compression everywhere on the go. Therefore, we have developed a portable compression device (PCD) that is small in size and convenient to use (Fig. 7.9). The PCD comprises three main constituents including a portable
actuation unit to provide hydraulic power, a smart compression sleeve to produce compression motion, and a user-friendly smartphone application for users to give commands.

Figure 7.9. Portable compression device composition.

7.5.1. Portable actuation unit

We developed the portable actuation unit to drive the smart compression sleeve without the need for a bulky central control unit (computer, power source, and data acquisition board) (Fig. 7.9). The portable actuation unit consisted of a stepper motor (Nema 11, Stepperonline, USA) with a 100 mm lead screw that drives a Luer-lok$^\text{TM}$ 1-mL syringe (BD, USA), a 3-cell lithium-ion polymer (LiPo) battery with a capacity of 800 mAh (Frontline Hobbies, Australia), and an in-house development printed circuit board (PCB). The PCB was housing a microcontroller ESP32-C3-MINI-1 (Espressif System, China) that supports Bluetooth Low Energy (BLE), a stepper motor controller integrated circuit (IC) DRV8825 (Texas Instruments, USA), and other functional assemblies such as a 3.3 V voltage conversion and a programming cluster.
We chose Bluetooth instead of Wi-Fi as the wireless communication protocol for the actuation unit because it would be less restrictive for users with an active lifestyle. Users would feel more comfortable using the device while out and about, away from a home access point. Wi-Fi’s benefits such as long-range communication and high bandwidth are not necessary for this device. In contrast to other Bluetooth protocols that are optimized for streaming data continuously, BLE is particularly suited for infrequently communicating small packets of data, as in the case of our PCD.

We programmed the PCB using the Arduino Software (also known as IDE) (Arduino, USA) with available packages for BLE connectivity, WS2812 addressing, and stepper motor acceleration. The software receives and processes command strings from BLE and then send commands to the stepper motor to regulate the compression sleeve elongation accordingly. Similarly, the software can write strings back to BLE which can be read by the smartphone application (e.g. remaining battery percentage).

All the actuation unit components were enclosed inside a 3D-printed box made from polylactic acid (PLA) by a 3D printer (Ultimaker, Netherlands). The 3D-printed box has a dimension of 210 mm in length, 55 mm in width and 53 mm in height and was equipped with an adjustable waist strap to facilitate wearing. We placed a limit switch at the ball screw distal end to register and set up the plunger’s initial position each time the stepper motor starts. The fully assembled portable actuation unit weighs 480 g.

7.5.2. Smart compression sleeve

We fabricated a fabric-based hydraulic artificial muscle (FHAM) with a dimension of 158 mm in length, 52 mm in width and 6 mm in thickness (Fig. 7.9). The FHAM consisted of a single HAM (OD2.5 × L1060 mm) arranged in six equal lines and inserted inside channels of a one-way stretchable fabric (self-adhesive bandage, 3M, Australia). Two pieces of self-adhesive bandage were sewn together in the stretchable direction (length-wise) to create six channels for HAM arrangement. The self-adhesive bandage can endure up to 100% strain, which is well suited to HAM’s elongation. Two ends of the FHAM were sewn to velcro straps to facilitate fastening such as to form a smart compression sleeve that encircles the user’s limb. Detailed specifications of the FHAM are shown in Table 7.1.
7.5.3. Smartphone application

To control the smart compression sleeve motion easily, we developed a smartphone application using MIT App Inventor (MIT, USA). MIT App Inventor is a powerful development environment that uses a graphical user interface (GUI) in which users can create a smartphone application by dragging and dropping visual objects [53]. We utilized the BLE package on the MIT App Inventor to set up a simple Bluetooth communication protocol. Specifically, the microcontroller ESP32-C3-MINI-1 is initialized as a BLE server with a single service that has a single characteristic value to be read and written. The smartphone is set up as a BLE client which can request and write to the same characteristic value to establish communication.

We also created a friendly user interface for the smartphone application that allows users to scan, select, and connect to the portable actuation unit (Fig. 7.9). Once connected, users can control the actuation unit and subsequently the smart compression sleeve by touching the corresponding buttons or dragging a specific indicator on the smartphone screen. For example, the “Home” button commands the plunger to return to a designated position. Dragging the “Position” indicator will move the plunger to the corresponding position which subsequently elongates or contracts the smart compression sleeve. The “Oscillate” button gives sinusoidal motion to the plunger whose amplitude can be adjusted by dragging the “Oscillation Amplitude” indicator.

7.5.4. Demonstration

Through various tests to determine the optimal motor speed without skipping steps, we decided to use the maximum speed of 1200 steps/s with an acceleration of 2000 steps/s². The stepper motor has 200 steps/revolution with a 2 mm pitch lead screw, inferring the actuation unit can provide a maximum linear speed of 12 mm/s to the syringe plunger. Including acceleration and deceleration, the portable actuation unit covers its full stroke of 50 mm in 4.7 s. Using the Luer-lok™ 1-mL syringe (ratio 57.3 mm/mL), we supplied 0.87 mL water to the FHAM to produce 18% elongation (from 158 mm to 187 mm) (Fig. 7.10A). The FHAM’s elongation was continuously controlled by dragging the “Position” indicator in the smartphone application. The communication was happening in real-time with inconsiderable latency.

We also measured the compression force generated by the smart compression sleeve through a hoop stress experiment (Fig. 7.10B). The hoop (OD 60 mm) comprised two half-cylindrical shells with a load cell (Futek, USA) sandwiched between them. The smart compression sleeve was pressurized and wrapped around the hoop using its integrated velcro straps. The
compression force increased when input pressure was reduced. This allows an easy adjustment of compression force to suit the specific needs of users. With a loading pressure of 0.98 MPa, the sleeve generated a maximum compression force of 19.5 N. Maximum compression force can be adjusted by varying loading input pressure. For example, a loading pressure of 0.52 MPa and 0.78 MPa generated a compression force of 8 N and 16.4 N, respectively (Fig. 7.10B). These compression forces are the average values of five measurements.

![Graph showing compression force vs. input pressure](image)

**Figure 7.10.** Portable compression device demonstration. (A) Elongation in a straight configuration. (B) Compression force of the compression sleeve with respect to input hydraulic pressure. (C) Actual setup allows a user to control the compression effect through a smartphone.

We then demonstrated the compression motion of the PCD (Fig. 7.10C and Supporting Video). A user wore the portable actuation unit at his back using the waist strap. After establishing a Bluetooth connection between the actuation unit and the smartphone application, the user dragged the “Position” indicator to the right to elongate the FHAM. He then attached the FHAM to his right forearm and moved the “Position” indicator to the middle position to generate an initial compression force. Subsequently, the user touched the “Oscillate” button to provide the FHAM (smart compression sleeve) with sinusoidal motion which induced alternative compression and relaxation to the forearm, like massage. The compression force can be adjusted by dragging the “Oscillation Amplitude” indicator. The smart compression sleeve took around 5 s to complete a motion cycle with a peak-to-peak amplitude of the plunger.
of 30 mm. Note that higher oscillation amplitudes will lower the compression frequencies because more time is required to complete a motion cycle. The user could put on and take off the entire device (portable actuation unit and smart compression sleeve) by himself without the help of others.

Regarding battery life, the stepper driver DRV8825 was set to deliver a maximum current of 850 mA to the stepper motor Nema 11. This setup makes the motor draw a total of 210 mA at 12 V (or 2.52 W) while running. Our LiPo battery has a capacity of 800 mAh and a nominal voltage of 11.1 V, meaning it can provide a total energy of 8.88 Wh. Therefore, battery life while running constantly is around 3.5 hours, which was tested and confirmed practically. However, it is not practical that the device is constantly running at such a high frequency. Therefore, we conducted another test with an intermittent compression pattern, including 2.5 s compression, rest for 10 s, 2.5 s relaxation, and rest for 10 s (a total of 25 s per cycle). The motor consumed 0.86 W every hour under this running pattern, which means our current battery should last for more than 10 hours.

7.6. Multifunctional smart compression sleeves

Along with compression motion, the smart compression sleeve is capable of providing additional functions that elevate its usability and versatility in a wide range of applications. We adopt the concept of using textiles as substrates to attach electronic components in soft wearable technologies to develop multifunctional smart textiles. However, instead of being passive, our smart textiles are active and controllable substrates, resulting in “smarter” devices with compact sizes and multiple functions. The possibility to integrate extra functions into smart textiles is endless. In this study, however, we limit our proof-of-concept demonstration to several popular functions including contact force sensing, and a substrate for electronic devices.

7.6.1. Contact force sensing

We fabricated a soft, stretchable liquid metal microtubule sensor (LMMS) and incorporated it into a smart woven sheet (Fig. 7.11A). The smart woven sheet, has a dimension of L140 × W72 mm, being the WHAM2 embodiment made from a 914 mm long HAM (OD 3.2 mm) and acrylic yarns. The sheet can be worn as a compression sleeve using integrated velcro straps. Detailed specifications can be found in Table 7.1. The LMMS consisted of a silicone elastomer
microtubule (OD 1.68 mm, ID 0.76 mm, Saint-Gobain, France), filled with liquid metal–
eutectic gallium-indium (EGaIn, Sigma-Aldrich, USA). The LMMS layer was sandwiched
between the compression sleeve’s inner surface and the object’s exterior. Compression force
transferred from the compression sleeve to the object was reflected by a resistance change of
the LMMS because of the deformation of the EGaIn channel (Fig. 7.11B) [54-56]. Therefore,
the sensor can provide real-time compression force feedback.

We conducted hoop stress experiments to establish the pressure-compression force and
compression force-resistance change relationships. The experimental setup and schematic
diagram are shown in Fig. 7.11C where a load cell (Futek, USA) bridged a two-piece 3D-
printed hoop (OD 60 mm) to collect compression force generated by the sleeve. We pressurized
the compression sleeve to 0.9 MPa, then enclosed the hoop. Subsequently, we withdrew
pressure by a 0.1 Hz sinusoidal signal with an amplitude that kept the minimum pressure at 0.1
MPa. Experimental results revealed that the compression sleeve could generate a compression
force of 41.8 N (Fig. 7.11D). The pressure-compression force chart shows a small hysteresis
gap between the pressurizing and releasing profiles, which means there is insignificant energy
loss when the sleeve was switching from expanding to compressing and vice versa. The chart
also indicates a linear relationship between input pressure and output compression force of the
sleeve in the releasing phase (when reducing input pressure).

Figure 7.11E shows a proportional relationship between the sleeve compression force and
resistance change (ΔR/R₀) of the LMMS. When generating a maximum compression force
(41.8 N), the compression sleeve caused a resistance change of 4000%. There is a large dead
zone of compression force until 28.5 N before the LMMS resistance starts to change. The
compression force-resistance change chart also signifies a wide hysteresis gap between the
pressurizing and releasing profiles. Both the large dead zone and wide hysteresis gap can be
explained by the soft nature of the LMMS silicone microtubule, acrylic yarns, and the LMMS
arrangement when incorporated into the sleeve. The soft silicone tube itself has hysteresis when
undergoing iterative compression. The HAM within the sleeve compresses the LMMS silicone
microtubule and acrylic yarns before deforming the liquid metal channel, generating resistance
change. Furthermore, the LMMS experienced point-contacts (HAM filament crosses over
LMMS filament) rather than line- or surface-contacts, leading to the large dead zone and sharp
raising of resistance change.
Figure 7.11. Smart woven sheet with integrated contact sensor and electronic components. (A) Liquid metal microtubule sensor (LMMS) is arranged on one face of the woven sheet. (B) LMMS resistance increases under compression force. (C) Experimental setup and schematic diagram of hoop stress measurement. (D) Relationship between output compression force and input pressure. (E) Relationship between resistance change of the LMMS and compression force. (F, G) Incorporating twinkle lights into the smart woven sheet. (H, I) Elongating the sheet while the lights are turned on.
7.6.2. Active substrates for electronic components

Our smart compression sleeves can also be implemented as active substrates to incorporate other electronic components without compromising their functionality. We demonstrate here an example of integrating a string of micro light-emitting diodes (LEDs) or twinkle lights into the WHAM2 embodiment (Fig. 7.11F, G). We interlaced the string of twinkle lights into the sheet in the weft direction (same as the yarn direction) so that it does not impede the sheet’s elongation. Upon pressurization, the WHAM2 could induce 35% strain and elongate back and forth normally while the twinkle lights were flashing (Fig. 7.11H and Supporting Video). Also, the woven light string did not qualitatively affect the sheet flexibility (Fig. 7.11I).

7.7. Discussion and conclusion

This paper presented a smart woven textile constructed from a high length-per-diameter hydraulic artificial muscle by the weaving technique. The artificial muscle itself is created by inserting a silicone tube inside a coil spring, which was introduced in our previous study [12, 57]. The smart sheet expanded 34.3% of its initial length when receiving 0.52 mL hydraulic volume. It is comparable to other smart textiles such as a knitted sleeve made of shape memory alloy (SMA) with 36% strain [58] and a smart textile constructed from McKibben muscles with 7.1% strain [35]. Our smart textile exhibited nonlinear hysteresis behavior evidenced by a gap between the pressurizing and releasing profiles in its elongation-volume characteristic. The hysteresis increases with increasing input amplitude and frequency (Fig. 7.2).

The nonlinear hysteresis behavior of the smart sheet increases the complexity of control schemes for precise position control. Therefore, we have investigated and proposed a hysteresis model based on the simplified Bouc-Wen structure. The identified model showed remarkable performance in capturing the output motion of the smart woven textile. This model was then implemented into a feedforward control scheme based on inverse compensation. Improving on this, we have developed and implemented a self-tuning adaptive controller that can control the output elongation of the smart sheet without prior information about the system. Both control schemes succeed in minimizing the RMSE between the output and desired elongation signals, which was especially prevalent with low-amplitude input signals (Fig. 7.8). The adaptive controller was much more effective at minimizing error and showed a reduction of RMSE up to 90% compared with the normalized gain uncompensated system. The adaptive controller
was capable of rejecting low-amplitude disturbances but large changes to the system configuration would cause the system to enter an unstable state. Conversely, the feedforward controller offered less accuracy but was much simpler to implement and had no issues regarding stability. It is worth noting that the performance of both control schemes became worse with amplitude and frequency. This has been attributed to the model uncertainty and the speed limit of our current motorized linear slider.

We have also developed a portable smart compression device to provide compression motion to the human extremities to increase blood flow. It can potentially be used for patients suffering from swelling issues, poor blood circulation or venous disease. The compression device is equipped with a smart compression sleeve made of a hydraulic artificial muscle to generate compression force and a portable actuation unit to control the compression sleeve motion when receiving commands wirelessly from a smartphone application. The portable actuation unit is compact and lightweight so users can wear it comfortably around the waist. The compression force generated from the compression sleeve was easily, dynamically and in real-time adjusted by touching the corresponding button or indicator on the friendly user interface smartphone application. The smart compression sleeve generated a maximum compression force of 19.5 N. In the oscillation mode, the compression sleeve could provide a complete cycle of compression and decompression within 5 s, which is slower than the wire-fabric compression sleeve but faster than SMA-based garments. Regarding power consumption, our 52 mm wide compression sleeve drew 2.52 W when running, which is incredibly low compared to the 27 W consumption of a 38 mm wide SMA compression garment developed by Holschuh et al. [59] or another prototype made by Golgoun et al. that drew 7 W for a 60 mm wide SMA garment [60]. Bluetooth communication between the smartphone application and the actuation unit was error-free in real-time with negligible latency.

Furthermore, the versatility of our smart textiles enabled the integration of other electronic components into their structures for additional functions. We demonstrated the integration of a filament liquid metal microtubule sensor that played the role of a contact force sensor. The resistance change of the integrated sensor was proportional to the compression force of the substrate smart compression sleeve. Thus, this resistance signal can potentially be fed back to the controller for precision control of the compression force on the user's limb. In another demonstration, we showed the integration of a string of twinkle lights into the entire smart woven sheet, where the oscillating elongation of the sheet did not affect the twinkle light function and vice versa. The demonstration might pave the way for more research into valuable
features such as thermistors, heating strips, thermoelectric coolers (Peltier), and other types of contact force sensors.

The development of smart woven sheets, control strategies and portable compression devices in this work has some drawbacks that need to be addressed in future work. Different sizes and materials of silicone tubes should be investigated to alleviate the effect of hysteresis behavior. The limited velocity of our current motorized linear slider hindered tests of high amplitude input signals with frequencies above 0.05 Hz. Therefore, replacing the slider with a faster one would improve the tracking performance of both the feedforward and adaptive controller as the system configuration would remain the same for a larger input frequency range and the adaptive controller would be able to respond to disturbances much faster. It is worth investigating the combination of feedforward and feedback controllers for high precision control requirements because this combined control strategy outperforms the feedback-only scheme and has even lower tracking error and higher stability [38]. Regarding the portable smart compression device, a faster and more powerful motor is required to speed up the compression cycle because rapid compression motion is more effective in treating venous leg ulcers than standard intermittent compression patterns, as shown by Nikolovska et al. [61]. The more powerful motor also promotes the development of larger compression sleeves to provide more compression force and cover a larger compression area. Finally, there is room significant for the miniaturization of the portable actuation unit to enhance user comfort.

In conclusion, we have introduced the development of a smart textile from woven hydraulic artificial muscle. We proposed a nonlinear hysteresis model to capture its hysteresis behavior. Subsequently, feedforward and adaptive controllers were developed to track its output elongation. We have also implemented the smart textile into a portable smart compression sleeve for compression therapy for people with poor blood circulation or venous disease. The compression sleeve can be controlled wirelessly via a smartphone application. We also demonstrated the feasibility of integrating a force sensor and other electronic components into the smart textile for additional functions. We expect this work will expand knowledge and contribute to the growth of the robotic research community and potentially benefit society.
Chapter 7. Modeling and control of woven hydraulic artificial muscles

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This chapter introduces another application of the proposed artificial muscle, which is a robotic cardiac compression device to assist in the treatment of heart failure. Heart failure is an emerging epidemic worldwide. It is a condition in which the heart does not pump sufficient blood to meet the body’s metabolic needs. The robotic cardiac compression device is designed to surround the epicardium to actively provide compression to the heart to assist the native heart in the pumping of blood out of the ventricles. The proposed device is made from a special configuration of artificial muscle filaments, allowing it to generate three-dimensional movements that mimic the natural motion of the heart. The spatial motion including radial, axial and torsional movements as well as the pumping performance of the device is experimentally measured. A mathematical model based on experimental calibration is developed to describe the three-dimensional motion of the device. In addition, an artificial pericardium is created as a sensing element and damping layer to relieve local stress and distribute compression from the device to the heart. Another configuration of the robotic cardiac compression device with an additional artificial muscle is fabricated to improve performance. Several variants of the proposed device are deployed to a porcine heart to demonstrate the device’s functionality.

This work has been submitted:

8.1. Abstract

Heart failure (HF) is a chronic condition that afflicts an estimated 64.3 million people worldwide. It occurs when the heart muscle cannot pump adequate blood to maintain the body’s normal function. People with HF often experience fatigue, shortness of breath and fluid buildup in the body. Treatment options for HF include physiotherapy, medication, mechanical heart support, heart surgery and heart transplantation depending on the disease’s progression. Effective treatments for HF are highly sought after. The medical robotics field has provided several types of mechanical devices to assist in the treatment of HF, including ventricular assist devices, passive ventricular constraint devices and direct cardiac compression devices. Despite advances, ventricular assist devices face the problem of device-blood contact, passive ventricular constraint devices have modest therapeutic efficacy and current direct cardiac compression devices are either bulky or unable to mimic the natural motion of the heart. This study introduces a robotic cardiac compression device made of artificial muscle filaments to actively generate compression to help the heart pump blood. The robotic cardiac compression device can produce spatial motion including radial, axial and torsional movements. The three-dimensional motion and performance of the proposed device are measured experimentally. An empirical model is developed to describe the motion of the device. Additionally, an artificial pericardium is fabricated to work as a sensing feature and provide soft contact between the device and the heart. The proposed device structure is highly customizable, meaning it can be modified to improve performance. Several variants of the proposed device are investigated experimentally using an ex-vivo porcine heart model to evaluate the device’s functionality. This proof-of-concept study is estimated to inspire further improvement and development of cardiac assist devices in the future.

8.2. Introduction

The heart is one of the most important organs in the vertebrate body. The heart and blood vessels make up the cardiovascular system that circulates blood throughout the body to deliver oxygen and nutrients, and remove carbon dioxide and other wastes [1]. The heart muscle is powerful and robust, working relentlessly to maintain the proper functions of the entire body. Heart failure (HF) is an emerging epidemic that affects approximately 64.3 million people worldwide [2]. HF is a condition in which the heart cannot supply sufficient blood to the body like a normal heart [3]. HF symptoms may include a rapid or irregular heartbeat, shortness of
breath, chest pain and swelling in the legs from fluid buildup. The most common causes of HF are myocardial infarction (heart attack) and coronary heart disease. Other diseases that may lead to HF include diabetes, high blood pressure, arrhythmias (irregular heart rhythm) and cardiomyopathy (heart muscle disease) [4].

HF is a chronic condition that often requires lifelong treatment to alleviate symptoms and improve quality of life. Treatment of HF usually involves lifestyle changes, cardiac rehabilitation and medications for early-stage HF [5-7]. For advanced stages or severe HF, surgical intervention or even heart transplantation is recommended. Some common surgical procedures include coronary artery bypass graft (CABG) for people with coronary heart disease, catheter ablation to treat abnormal heart rhythms and heart valve surgery to repair or replace malfunctioning heart valves [8]. In some cases of extremely severe HF, a heart transplant is recommended [9]. However, not everyone is eligible for a heart transplant depending on their health status. Also, the number of donated hearts is very limited.

There is a huge demand for an effective method to treat HF. Medical robotics has provided a class of implantable devices that support the pumping function of the heart. Some devices are basically mechanical pumps that take over the heart’s function in circulating blood while others surround the heart to provide additional compression. These devices are recommended for the treatment of advanced HF, either as bridging therapy for patients awaiting a heart transplant or as destination therapy. There are broadly three types of mechanical implantable devices for the treatment of HF.

The first type is ventricular assist devices (VADs), also known as mechanical circulatory support devices. These devices are mechanical pumps deployed into the circulatory system to move blood throughout the body [10]. Most VADs are implanted in the left ventricle (LV) to circulate oxygen-rich blood from the LV to the body through the systemic circulation and are referred to as left VADs or LVADs. Less commonly, VADs are used for the right ventricle (RV) to pump oxygen-poor blood from the RV to the lungs through the pulmonary circulation and are called right VADs or RVADs. There are also biventricular assist devices (BiVADs) to assist both ventricles. A VAD typically comprises a mechanical pump, an inflow cannula, an outflow graft and a drive system. The mechanical pump can be centrifugal or axial flow for blood circulation. It is usually implanted directly into the ventricles to receive blood through a short inflow cannula. However, some designs prefer to place the pump outside the body with a long inflow cannula connected to the ventricles. The outflow graft is to return blood from the
pump back to the corresponding blood vessels. The drive system is located outside the body and is connected to the pump via a driveline cable.

Some well-known VADs are Heart Assist 5 [11, 12], Jarvik 2000 [13, 14], HeartMate 3 [15, 16], HeartWare and TandemHeart. The HeartWare HVAD System (Medtronic, USA) is a mechanical implantable cardiac assist device that uses a centrifugal flow pump implanted directly into the bottom of the LV (the apex). The pump circulates blood from the LV to an outflow cannula connected to the aorta [17]. Several clinical studies have shown HeartWare’s capability to provide adequate blood flow, thereby improving symptoms of HF [18-20]. However, Medtronic decided to stop the distribution and sale of the HVAD System in Jun 2021 due to risks of neurological adverse events, mortality and potential failure to restart the device [21]. TandemHeart (LivaNova, USA) is an extracorporeal VAD, which means the pump is placed outside the patient’s body. The TandemHeart features a centrifugal flow pump to circulate blood, an inflow cannula inserted into the femoral vein, puncturing the septum to access the left atrium and an outflow cannula inserted into the femoral artery [22, 23].

In a breakthrough development, BiVACOR Inc., USA has created a total artificial heart (TAH) that completely takes over the total function of the patient’s native heart [24, 25]. The BiVACOR TAH is designed for long-term implantation in patients with end-stage HF. The device features a centrifugal pump with a double-sided impeller to provide simultaneous perfusion to the systemic and pulmonary circulations. The device and its smart controller enable dynamic adaptation to changes in the patient’s physiology and are capable of providing a powerful cardiac output of up to 12 L/m. Treatment of HF with VADs helps alleviate symptoms, enhance heart function and improve quality of life [26]. However, VADs have hard components that come into direct contact with blood, causing potential adverse events including infection, bleeding and device thrombosis [10, 27-29]. Furthermore, patients are required to take blood thinners during treatment to prevent blood clot formation.

The second type of mechanical device for the treatment of HF is passive ventricular constraint devices (PVCDs). These devices are constructed from passive elastic materials to limit cardiac enlargement [30]. PVCDs are intended to enclose the epicardium to decrease stress on ventricular walls and reduce heart muscle stretch. PVCDs are recommended for people with LV remodeling and dilated cardiomyopathy (weak and enlarged heart muscle) to prevent further dilation [31-34]. The HeartNet ventricular constraint device (Paracor Medical, USA) is a flexible band made from nitinol wires coated with silicone. The band is wrapped around the
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Heart circumference to provide pressure to the epicardium [31]. The HeartNet with a high rate of transplant success (98%), has the potential to reverse LV remodeling [35, 36]. Acorn CorCap cardiac support device (Acorn Cardiovascular, USA) is a polyester mesh that encloses the entire heart surface from base to apex to limit cardiac dilation [31]. Clinical results have demonstrated the long-term safety of CorCap implantation and the potential for LV reverse remodeling [37]. Instead of making a fixed-size PVCD, Ghanta et al. developed an adjustable ventricular restraint device. The device has a cup shape with two thin layers made from medical-grade polyurethane [38]. The space between two layers can be inflated or deflated to vary the pressure applied to the heart’s surface. The restraint level can be adjusted periodically to improve treatment outcomes.

PVCDs surround the epicardium and are not in contact with circulating blood. Therefore, using PVCDs in the treatment of HF avoids complications of blood clot formation and bleeding. However, the therapeutic efficacy in improving symptoms of HF is limited due to their principle of passive constraint rather than active compression [39, 40]. Therefore, PVCDs are suitable for the treatment of early-stage HF to prevent further ventricular dilation, thus slowing or ceasing disease progression.

The third type of mechanical device for HF treatment is direct cardiac compression devices (DCCDs). These devices typically have conical shapes, allowing easy attachment to the epicardium. Surgical sutures or suction cups are often required to maintain contact between the device and the heart. DCCDs are designed to actively provide compression to assist the heart during systole to pump blood out of the ventricles [41]. In cases where good adhesion is established between the device and the heart, the device can support diastole to actively fill the ventricles with blood. To synchronise the compression and expansion of the device with the systole and diastole of the heart, an electrocardiogram (ECG) is required in the control system. Treatment of HF with DCCDs eliminates complications from blood contact with the device, which remains a major problem for VADs [9, 29]. In addition, DCCDs provide active movements rather than passive constraints like PVCDs, which means that DCCDs have the potential to deliver greater therapeutic effects [28, 42].

The development of DCCDs has a long history, dating back to 1956 with the pneumo-massage introduced by Bencini and Parola to massage the heart [43]. Later, the Anstadt cup for cardiac rehabilitation to increase blood pressure and cardiac output appeared in 1965 [44-46]. These promising results inspired many subsequent developments. AbioBooster cardiac assist device
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(AbioMed, USA) is designed to surround the heart for long-term assistance. The device features a number of silicone-coated inflatable polyurethane tubes arranged in a cylindrical shape to enclose the epicardium [47]. When positive pneumatic pressure is applied, these tubes inflate, generating compression to support systole. Conversely, these tubes are deflated by removing the input pressure to accommodate diastole. One study showed that the AbioBooster could produce an aortic pressure of 115 mmHg and a blood flow rate of 6.5 l/min [48]. In a slightly different approach, HeartPatch (Heart Assist Technologies, Australia) has two independent inflatable silicone patch components that are attached to the two ventricles with silicone adhesive [49]. The patches are pneumatically driven to provide compression to the ventricles. This design allows the two patch chambers to be controlled independently so that compression intensity and frequency can be customized for each ventricle [28, 50]. An ECG is used to synchronize the movement of the device with the heart. Potential limitations of the HeartPatch include delamination of the patches from the cardiac surface and the rupture of heart muscle [41].

Soft artificial muscles (SAMs) have emerged as a potential actuation mechanism for many robotics and medical applications such as wearable assistive devices, surgical robots and smart garments [51-54]. Their implementation in cardiac assist devices has been explored recently with impressive achievements [9, 41]. The softness, conformability and controllability of SAMs are desirable to make gentle contact with the heart surface as well as providing active compression. Roche et al. developed a soft robotic sleeve that wraps around the epicardium to assist compression [55, 56]. The device has a cup shape made of two layers of pneumatic artificial muscles (PAMs) coated with silicone. One layer has the PAMs arranged in the circumferential direction to produce radial compression while the PAMs of the other layer are aligned axially with a twist to induce twist motion. The two layers of the soft robotic sleeve can be controlled independently, enabling the implementation of complex control strategies to mimic the heart’s natural movements. However, the soft robotic sleeve is relatively large with a thick wall, which is not favorable for implantation. A complex control system and a bulky air source are required to run the device. Kongahage et al. introduced a new extracorporeal VAD made from electrothermal actuators coated with silicone that could produce pulsatility instead of the continuous flow generated by conventional VADs [57]. The author also presented the design of a DCCD using the proposed soft actuators to enclose and assist heart compression.

CorInnova DCCD (CorInnova, USA) is a double-walled cup-shaped device for the treatment of acute HF [58]. A series of nitinol wires separate the two silicone walls (or layers). The cavity
between the two layers is filled with compressed air to create compression during systole. The space between the inner layer and the heart is filled with liquid to improve the interaction between the device and the heart [59]. The device is capable of multifunctional assistance including active compression and passive constraint [60]. Interestingly, the CorInnova device can be implanted with minimally invasive surgery, causing less postoperative pain and shorter hospital stays for patients. However, the CorInnova device has limited motion (squeeze by inflating the air chamber) so it cannot replicate the natural movements of the heart. Therefore, the CorInnova device may not be suitable for long-term implantation.

This study introduces a robotic cardiac compression device (RCCD), which is a DCCD for the treatment of advanced HF. The primary drive of this study was the clinical need for a simple and effective cardiac assist device. We also want to address some of the aforementioned limitations of current DCCDs. The proposed RCCD is constructed from our previously developed fluid-driven artificial muscle filament, which is flexible, responsive and has a high length-to-diameter ratio and high elongation [61]. A special arrangement of artificial muscle filaments allows the RCCD to produce spatial motion including radial, axial and torsional, mimicking the natural motion of the heart. We measure several aspects of RCCD performance such as stroke volume, flow rate, compression force and output pressure. An empirical model is proposed to describe the three-dimensional (3D) motion of the RCCD. We create an artificial pericardium and integrate it inside the RCCD to relieve stress on the heart surface and also use it as a sensing component. Furthermore, we introduce a modified RCCD that has an extra artificial muscle filament to enhance the original RCCD performance. Finally, we demonstrate RCCD performance on a dead porcine heart.

8.3. Design and fabrication of a robotic cardiac compression device

This section introduces the design concept, working principle and fabrication process of the RCCD – a type of direct cardiac compression device that surrounds the heart to actively assist in the treatment of HF. We aim to create an RCCD that replicates 3D cardiac motion (radial, axial and twist in 3D space) using a special configuration of our developed soft artificial muscles. The RCCD is a semi-spheroidal shell consisting of an inner layer made of silicone elastomer and an outer layer made of specially arranged artificial muscle filaments (AMFs) (Fig. 8.1A). The inner layer is a soft, thin, low-Young’s modulus silicone elastomer to stabilise the outer layer structure as well as provide soft contact with the epicardium. The outer layer is
formed by two AMFs, including a compress-AMF inside and adjacent to a twist-AMF. The compress-AMF is helically wrapped around the silicon layer to induce radial motion. The twist-AMF is folded into multiple segments arranged along the RCCD axis of symmetry with a predefined twist to provide axial and twist movements.

Figure 8.1. Robotic cardiac compression device (RCCD). (A) Design concept. RCCD is a semi-spheroidal shell made of two layers of specially arranged artificial muscle filaments (AMFs). (B) Working principle. RCCD compresses the epicardial surface during systole and expands during diastole. (C) RCCD prototype.
Chapter 8. Robotic cardiac compression device

The compress-AMF and twist-AMF are thin, long and flexible fluid-driven artificial muscles made by inserting a silicone tube inside an extension coil spring. One end of the AMF is blocked while the other end is connected to a hydraulic source via a flexible fluid transmission tube. This simple fabrication method of insertion using commercially available components allows the AMF to have a high length-to-diameter ratio, high reliability and repeatability, enabling low-cost mass production. When receiving positive hydraulic pressure, the AMF extends longitudinally due to the radial constraint of the coil spring on the silicone tube. Conversely, when the pressure is released, the AMF contracts and returns to its original length. The AMF is responsive, durable, has high elongation and high energy efficiency that can be deployed in many robotic and medical applications such as wearable robotic gloves, smart surgical sutures, smart textiles, compression garments and flexible 3D bioprinters [61-65].

The special configuration of AMFs in the RCCD allows the transformation of the AMFs’ longitudinal motion into the RCCD spatial motion that mimics the natural heart motion. Specifically, when pressure is applied, the compress-AMF’s elongation is transformed into the RCCD’s radial expansion while the twist-AMF’s elongation is converted into the RCCD’s axial expansion and twisting motion in one direction. This pressurising phase of the RCCD corresponds to the diastolic phase of the heart when the heart muscle relaxes to allow blood to passively fill the atria and partially fill the ventricles. Subsequently, when pressure is withdrawn, the contraction motion of the two AMFs generates the RCCD’s radial and axial contraction and twisting motion in the reserve direction. This releasing phase of RCCD coincides with systole when the heart contracts to pump blood out of the ventricles (Fig. 8.1B). Thereby, two working phases of RCCD complete the cardiac cycle.

We fabricated an RCCD prototype using the following process. First, we created a thin silicone layer from Ecoflex 00-30 (Smooth-On, USA) using the molding technique. A set of molds made of polylactic acid (PLA) were manufactured using a 3D printer (Ultimaker, Netherlands). The silicone layer prototype had a semi-spheroidal shape with a base diameter of 76 mm, a height of 78 mm and a thickness of 1.5 mm. Second, we made two AMFs with the same specifications (e.g. outer diameter of 2.5 mm) except for their lengths, where the compress-AMF had a length of 1000 mm and the twist-AMF was 940 mm long. Both AMFs were connected to the same flexible fluid transmission tube via a T-connector so they could be pressurised at the same time. Detailed specifications of the prototypes can be found in Table 8.1. Third, we pressurised the two AMFs to 35% strain (input pressure around 1.2 MPa), then helically wrapped the compress-AMF around the silicone layer (with its inner mold attached)
and arranged the twist-AMF on top of the compress-AMF. The intersection points of the two AMFs were secured and attached to the silicone layer by elastic strings. It is worth noting that the silicone layer was molded with designated grooves on its exterior to facilitate the arrangement of AMFs. Last, we removed the inner mold to obtain the RCCD prototype (Fig. 8.1C). The result was a base diameter of 68 mm, a height of 72 mm and a thickness of 7 mm at the releasing phase.

**Table 8.1. Specifications of prototypes**

<table>
<thead>
<tr>
<th>Artificial muscle filament (AMF)</th>
<th>Coil spring</th>
<th>Silicone tube</th>
</tr>
</thead>
<tbody>
<tr>
<td>Carbon steel (AliExpress, China)</td>
<td>Silicon elastomer (Saint-Gobain, France)</td>
<td></td>
</tr>
<tr>
<td>OD: 2.5 mm</td>
<td>OD: 2.2 mm</td>
<td></td>
</tr>
<tr>
<td>ID: 1.9 mm</td>
<td>ID: 1 mm</td>
<td></td>
</tr>
<tr>
<td>k: 21.4 N/m</td>
<td>E (100%): 1.586 MPa</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Robotic cardiac compression device (RCCD)</th>
<th>Configuration A</th>
<th>Configuration B</th>
</tr>
</thead>
<tbody>
<tr>
<td>Compress-AMF: L 1000 mm</td>
<td>Configuration A + additional compress-AMF: L 755 mm</td>
<td></td>
</tr>
<tr>
<td>Twist-AMF: L 940 mm</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Silicone layer: Ecoflex 00-30 (Smooth-On, USA), base diameter 76 mm, height 78 mm, thickness 1.5 mm</td>
<td></td>
<td></td>
</tr>
<tr>
<td>RCCD size: base diameter 68 mm, height 72 mm, thickness 7 mm</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Artificial pericardium</th>
<th>Outer layer</th>
<th>Inner layer</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ecoflex 00-30 and satin fabric</td>
<td>Ecoflex 00-30 and satin fabric</td>
<td></td>
</tr>
<tr>
<td>Height: 78 mm</td>
<td>Height: 72 mm</td>
<td></td>
</tr>
<tr>
<td>Base diameter: 72 mm</td>
<td>Base diameter: 68 mm</td>
<td></td>
</tr>
<tr>
<td>Thickness: 1.6 mm</td>
<td>Thickness: 1.6 mm</td>
<td></td>
</tr>
</tbody>
</table>

OD: outer diameter, ID: inner diameter, L: length, k: spring constant, E: Young’s modulus
8.4. Experimental characterisation

8.4.1. Spatial motion of RCCD

The RCCD is designed to induce spatial motion including radial expansion $\varepsilon_r$, axial expansion $\varepsilon_z$ and twist angle change $\delta\omega$ when pressurising and vice versa (Fig. 8.2A). We built an experimental platform to capture RCCD motion and measure the shape change corresponding with input pressure (Fig. 8.2B). The platform consisted of a fixture to which the base of the RCCD prototype (Fig. 8.1C) was attached, so that its apex (or tip) could move freely, two side cameras (Nikon, Japan and Nano Shield, Australia) $90^\circ$ apart were used to capture RCCD dimension change in the radial and axial directions and a bottom camera (Nano Shield, Australia) was used to capture the twist angle change. Note that one side camera can do the job but averaging two cameras helps to minimise errors caused by the asymmetrical shape of the RCCD prototype during fabrication. A motorised linear slider (Zaber, Canada) was used to generate movements of a 3-mL syringe (BD Biosciences, Canada) to pump fluid (degassed water) to the RCCD.

We supplied the motorised linear slider with a step signal comprising ten steps forward with 4.4 mm increments and ten steps backward with 4.4 mm decrements, with each step 3 s apart. The slider speed was 10 mm/s. The shape change of the RCCD at each step was captured by the three cameras and subsequently underwent image processing to calculate dimension changes. A pressure sensor (Honeywell, USA) was connected to the fluid transmission tube to measure input pressure. The experiment was run five times. Experimental data are presented as mean $\pm$ standard deviation.

Figures 8.2C, D show the side view and bottom view of the RCCD at different input pressures. As the input pressure increased, the RCCD expanded in radial and axial directions as well as twisted clockwise. Conversely, when the input pressure decreased, the RCCD radially and axially contracted and twisted counterclockwise. The magnitudes of the radial, axial and twisting movements varied slightly at different latitudes with an increase from base to apex. The base of the RCCD experienced the least dimension change because it was tightened to the fixture. We present in Fig. 8.2E-G the experimental results which are the relationships between three output parameters (axial expansion, radial expansion and twist angle change) and the RCCD input pressure at the latitude of the compress-AMF starting point (near the apex, marked with a cyan dot in Fig 8.2C). With an input pressure of 1.13 MPa, the RCCD induces an average radial expansion of 34.5%, an axial expansion of 14.5% and a twist angle change of 32.2°. All
three charts show a clear nonlinear hysteresis relationship between the output parameters and input pressure, where a gap exists between the pressurising and releasing profiles. Furthermore, the charts revealed the fast responsiveness of the RCCD evidenced by the absence of backlash, meaning that any change in input pressure causes a corresponding change in the RCCD motion.

**Figure 8.2.** Characteristics of robotic cardiac compression device (RCCD). (A) Spatial motion illustration. (B) Experimental setup using three cameras for motion capture. (C) Side view shows axial and radial expansion at different input pressure. (D) Bottom view shows the change of twist angle. (E-G) Experimental results show the relationship between input pressure and axial expansion, radial expansion and twist angle change.
8.4.2. Performance of RCCD

We set up another testing platform to measure the performance of the RCCD in terms of stroke volume, flow rate and output pressure (Fig. 8.3A). The RCCD prototype enclosed a silicone rubber univentricle and was attached to a fixture. A flexible tube connected the univentricle to a reservoir through an ultrasonic flowmeter (Atrato, Titan Enterprises, UK). This connecting tube served as the inlet as well as the outlet of the univentricle. The reservoir was raised approximately 350 mm above the RCCD fixture to create initial hydraulic pressure. The ultrasonic flowmeter is capable of measuring instantaneous flow rate and cumulative volume of liquid flowing through its measuring channel. A pressure sensor was located on the connecting tube just 50 mm above the univentricle to measure output pressure. Similar to the previous experiment, the RCCD was actuated by a motorised linear slider via a 3-mL syringe with a pressure sensor attached to the input fluid transmission tube. We controlled the motorised linear slider using sinusoidal signals with an amplitude of 22 mm at various frequencies to generate RCCD motion. Note that we have limited the amplitude to 22 mm so that the input pressure does not exceed 1.2 MPa to avoid damage to the RCCD prototype.

The RCCD produced an average stroke volume of 107.6 mL at 0.1 Hz (or a rate of 6 beats per minute (bpm)), which equated to an average flow rate of 0.65 L/min (Fig. 8.3B). At the higher speeds of 12 bpm (0.2 Hz) and 15 bpm (0.25 Hz), the RCCD achieved average stroke volumes of 82 and 70 mL, which were equivalent to average flow rates of 0.98 and 1.05 L/min, respectively. Figure 8.3C shows the relationship between instantaneous flow rate and input pressure at 0.2 Hz. When the input pressure increased, the RCCD expanded, allowing water from the reservoir to flow down to fill the univentricle with a flow rate that could reach a maximum of 2.5 L/min. Subsequently, when the input pressure decreased, the RCCD contracted to pump water from the univentricle to the reservoir with a maximum instantaneous flow rate of 2.8 L/min. The output pressure (univentricular pressure) is inversely proportional to the input pressure (Fig. 8.3D). The output pressure reached a minimum of 16.3 mmHg when the RCCD fully expanded at an input pressure of 1.2 MPa, whereas when the RCCD completely compressed, it produced a maximum output pressure of 50 mmHg. A hysteresis gap exists between the pressurising and releasing phases of the input-output pressure chart.

We also tested the compression force and blocked output pressure generated by the RCCD. We used a two-piece hoop with a load cell (Futek, USA) sandwiched between them to collect compression force (Fig. 8.3E). The hoop shape resembles the RCCD shape for better contact.
Figure 8.3. Performance of robotic cardiac compression device (RCCD). (A) Experimental setup to measure flow rate and output pressure. (B) Stroke volume and average flow rate of RCCD at different speeds. (C) Flow rate at 0.2 Hz. (D) Output pressure at 0.2 Hz. (E) Compression force measurement setup. (F) Force-pressure hysteresis at different loading pressures. (G) Maximum force at different loading pressures. (H) Relationship between blocked output pressure and input pressure. (I) Maximum blocked output pressure at different loading pressures. bpm, beats per minute.

We pressurised the RCCD to a certain expansion before wearing it on the hoop and then reduced the input pressure using a 0.1 Hz sinusoidal signal with an amplitude that kept the minimum input pressure at 0.1 MPa. Three different loading pressures were used to accommodate three different sizes of hoop. The compression force has an inversely
proportional relationship with the input pressure, in which the compression force is almost zero at the loading pressure and peaks when the input pressure drops to the lowest level (0.1 MPa) (Fig. 8.3F). Figure 8.3F also reveals a modest hysteresis between the pressurising and releasing phases of the pressure-force relationship. Three hysteresis profiles, corresponding to three loading pressures, share similar hysteresis patterns. With the loading pressure of 0.6, 0.8 and 1 MPa, when the input pressure was withdrawn, the RCCD achieved a maximum compression force of 10.5, 18.7 and 25.3 N respectively (Fig. 8.3G).

The experimental setup to measure blocked output pressure was similar to that shown in Fig. 8.3A, but the univentricular outlet was blocked instead of connecting to the reservoir. We pressurised the RCCD with a certain loading pressure and had the univentricle fully filled with water, then blocked the outlet. Similar to the compression force test, we reduced the input pressure using a 0.1 Hz sinusoidal signal with an amplitude that kept the minimum input pressure at 0.1 MPa. The blocked output pressure was inversely proportional to the input pressure with a very small hysteresis gap (Fig. 8.3H). Corresponding to the loading pressure of 0.6, 0.8 and 1 MPa, the RCCD generated blocked output pressure of 59.7, 62.6 and 69.2 mmHg, respectively when the pressure was released (Fig. 8.3I).

8.5. Mathematical model

This section presents a mathematical model to describe the RCCD spatial motion including radial expansion, axial expansion and twist angle change corresponding to the input pressure. Due to the complexity of modeling hyperelastic components, we propose an empirical model based on the experimental calibration of AMF samples rather than theoretically building it from scratch. Our method includes an experimental investigation of the relationship between the AMF’s longitudinal elongation and the input pressure, also known as the calibration. Next, we study the geometric arrangement of the AMFs to establish the distribution of their longitudinal elongation to each constituent of the RCCD spatial motion. Last, the elongation distribution and calibration data are used to infer the relationship between the RCCD spatial motion and the input pressure.

Instead of calibrating the two long AMFs that formed the RCCD, which makes the experimental setup difficult, we conducted an experiment to calibrate a shorter AMF with the same specifications. We hypothesised that AMFs made of similar components have similar
pressure-elongation hysteresis profiles regardless of their lengths. The longitudinal elongation \( \varepsilon_l \) is defined by the ratio of the displacement \( \delta_l \) to the original length \( l_0 \) as \( \varepsilon_l = \delta_l/l_0 \). To validate this hypothesis, we created three AMF samples from similar silicone tubes and coil springs with an outer diameter of 2.5 mm and different lengths of 50, 100 and 150 mm (Fig. 8.4A). These AMFs have the same specifications as the two AMFs of the RCCD (Table 8.1). We supplied the three AMF samples with 0.1 Hz sinusoidal signals with different amplitudes that kept the maximum input pressure at 1.2 MPa and established the relationship between input volume, pressure and output elongation. It is intuitive that a longer AMF required a larger input volume to reach a certain threshold of pressure or elongation than a shorter one (Fig. 8.4B).

Interestingly, the pressure-elongation hysteresis profiles of the three AMFs almost overlapped with each other with a maximum standard deviation of 0.53% (Fig. 8.4C). The AMFs achieved an elongation of 36.8 ± 0.25% when an input pressure of 1.2 MPa was applied. The experimental results in Fig. 8.4C validated our hypothesis about the similarity of pressure-elongation hysteresis of AMFs with different lengths.

A nonlinear hysteresis model can be used to describe the pressure-elongation hysteresis profiles in Fig. 8.4C [66]. However, to facilitate the calculation in later steps, we propose a simpler approach, which is to use two different polynomial fitting curves to represent the pressurising and releasing phases. The fitting curves are governed by cubic polynomial functions with the coefficients shown in Table 8.2. The fitting curves are shown in Fig. 8.4C with the root mean square error (RMSE) of the pressurising and releasing phases being 0.33% and 0.17% respectively compared to the average elongation of the three AMFs.

Figure 8.4D illustrates the geometric arrangement of the compress-AMF and a segment of the twist-AMF to form the RCCD. The RCCD is the bottom half of a spheroid (also known as an ellipsoid of revolution) which can be described by the following parametric equation.

\[
\begin{align*}
  x &= a_0 \sin \theta \cos \varphi, \\
  y &= a_0 \sin \theta \sin \varphi, \\
  z &= c_0 \cos \theta
\end{align*}
\]

(1)

where the semi-axis \( a_0 \) is the equatorial radius of the spheroid, the semi-axis \( c_0 \) is the distance between the equatorial plane and the pole along the symmetry axis, \( \theta \) is the polar angle and \( \varphi \) is the azimuth angle.

The two AMFs resemble 3D helices surrounding the spheroid. A 3D helical spline on a spheroid can be expressed by Eq. (2).
where $\omega$ is the twist angle. The compress-AMF has $\theta \in [0.5\pi, 0.9\pi]$ and $\omega_c \in [0, 9\pi]$ while the twist-AMF has $\theta \in [0.5\pi, 0.9\pi]$ and $\omega_t \in [0, 0.32\pi]$.

To facilitate the comparison between the proposed model and the experiment, we choose to investigate the starting point of the two AMFs (point A in Fig. 8.4D), whose latitude is similar to the cyan dot shown in Fig. 8.2C. This starting point has a polar angle $\theta = 0.9\pi \text{ rad}$, a radius (distance to the symmetry axis) $r_0 = a_0 \sin \theta$ and an altitude (distance to the equatorial plane) $h_0 = c_0 |\cos \theta|$ (Fig. 8.4D).

**Figure 8.4.** Mathematical model of robotic cardiac compression device (RCCD). (A) Three AMF prototypes of different lengths are created for calibration. (B, C) Elongation-volume and elongation-pressure relationships of the three prototypes. (D) Geometric illustration of RCCD with spatial displacement. (E-G) Comparison between mathematical models and experimental results on radial expansion, axial expansion and twist angle change. AMF, artificial muscle filament; RMSE, root mean square error; P2PE, peak-to-peak error.
Table 8.2. Coefficients of polynomial fitting curves

<table>
<thead>
<tr>
<th>Function</th>
<th>Coefficients</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$k_1$</td>
</tr>
<tr>
<td>Pressurising phase (bottom)</td>
<td>-25.3277</td>
</tr>
<tr>
<td>Releasing phase (top)</td>
<td>-9.2204</td>
</tr>
</tbody>
</table>

The compress-AMF has an original length $l_{o,c}$ and wraps $n$ revolutions around the semi-spheroid. While all $n$ revolutions contribute to the radial expansion of the entire semi-spheroid, the expansion magnitude is attributed to a single revolution. The circumference of a revolution is called the effective length and can be roughly calculated as $l_c = l_{o,c} \eta_c / n$, where the coefficient $\eta_c < 1$ represents the efficiency of the compress-AMF in generating radial expansion. The coefficient $\eta_c$ also accounts for the different circumferences of each revolution due to the difference in latitude. The helix angle $\beta_c$ of the compress-AMF formed with the equatorial plane can be expressed by Eq. (3).

$$
\beta_c = \sin^{-1} \left( \frac{h_0}{l_{o,c} \eta_c} \right) = \sin^{-1} \left( \frac{c_0 |\cos \theta|}{l_{o,c} \eta_c} \right)
$$

(3)

The twist-AMF has an original length $l_{o,t}$ and is folded into $m$ segments. The effective length of the AMF to generate the magnitude of motion is $l_t = l_{o,t} \eta_t / m$, where $\eta_t < 1$ represents the efficiency of the twist-AMF in generating axial and twist motion. Both coefficients $\eta_c$ and $\eta_t$ indicate the ratio between cumulative lengths of active segments and the total length of the corresponding AMFs and are estimated during fabrication. Each AMF segment is twisted at a predefined angle $\omega_t$. The helix angle $\beta_t$ of the twist-AMF formed with the equatorial plane can be expressed by Eq. (4).

$$
\beta_t = \sin^{-1} \left( \frac{h_0}{l_t} \right) = \sin^{-1} \left( \frac{c_0 |\cos \theta|}{l_t} \right)
$$

(4)

From the calibration, we have the longitudinal elongation $\varepsilon_l$ of the AMF as a function of the input pressure $P$ i.e. $\varepsilon_l = f(P)$. From this, we can calculate the length-wise displacement of the compress-AMF $\delta_{l,c} = \varepsilon_l l_c$ and that of the twist-AMF $\delta_{l,t} = \varepsilon_l l_t$. Note that these displacements are calculated based on the effective length.
The compress-AMF contributes the majority of its motion in the radial direction while the twist-AMF generates axial and twist motion. The compress-AMF also induces a slight movement in the axial direction and the twist-AMF has a small contribution to radial motion. However, these movements are negligible and can be ignored to simplify the calculation. An important aspect of the AMF arrangement that needs to be considered is the interaction between the two AMFs. They are linked and tightened at their intersections, thereby impeding each other’s movements. We define two coefficients \( \gamma_c \) and \( \gamma_t \) to indicate this impediment phenomenon. The magnitude of these coefficients is the ratio between their respective AMF lengths and the sum of the two AMF lengths, as shown in Eqs. (5-6). Note that the term \( \sin \omega_t \) in Eq. (6) indicates the effect of \( \gamma_t \) specifically on the axial motion.

\[
\gamma_c = \frac{l_{o,c}}{l_{o,c} + l_{o,t}}; \quad \gamma_t = \frac{l_{o,t} \sin \omega_t}{l_{o,c} + l_{o,t}} \tag{5}, \tag{6}
\]

Radial displacement \( \delta_r \), axial displacement \( \delta_z \) and tangential displacement \( \delta_u \) at point A of the RCCD (Fig. 8.4D) are distributed from longitudinal displacements of the compress-AMF (\( \delta_{l,c} \)) and the twist-AMF (\( \delta_{l,t} \)) and are shown in Eq. (7).

\[
\begin{aligned}
\delta_r &= \frac{\delta_{l,c} \gamma_c \cos \beta_c}{2 \pi} \\
\delta_z &= \delta_{l,t} \gamma_t \sin \beta_t \\
\delta_u &= \delta_{l,t} \cos \beta_t
\end{aligned} \tag{7}
\]

Then, we can obtain the radial expansion \( \varepsilon_r \), axial expansion \( \varepsilon_z \) and twist angle change \( \delta_\omega \) by Eq. (8).

\[
\begin{aligned}
\varepsilon_r &= \delta_r / r_0 \\
\varepsilon_z &= \delta_z / h_0 \\
\delta_\omega &= \delta_u / r_0
\end{aligned} \tag{8}
\]

The mathematical model is then executed with input pressure ranging from 0 to 1.2 MPa, coefficients of polynomial fitting curves in Table 8.2 (based on experimental calibration of shorter AMF samples) and model parameters in Table 8.3 (based on the actual RCCD prototype). Figures 8.4E-G compare the results of the mathematical model and the experimental data of the RCCD spatial motion. Despite the simplification of the model, it shows very good performance when following the experimental data with a small RMSE of 0.95% for radial expansion, 0.37% for axial expansion and 1.6° for twist angle change. This empirical model is built upon calibration so it performed well in capturing the trend of
experimental data. However, it has the limitation of loosely following local changes, as evidenced by a relatively large peak-to-peak error (P2PE) of 3.81% for radial expansion, 1.02% for axial expansion and 3.8° for twist angle change.

Table 8.3. Model parameters

<table>
<thead>
<tr>
<th>Spheroid</th>
<th>Compress-AMF</th>
<th>Twist-AMF</th>
</tr>
</thead>
<tbody>
<tr>
<td>$a_0 = 34, \text{mm}$</td>
<td>$l_{o,c} = 1000, \text{mm}$</td>
<td>$l_{o,t} = 940, \text{mm}$</td>
</tr>
<tr>
<td>$c_0 = 72, \text{mm}$</td>
<td>$n = 6.5, \text{revolutions}$</td>
<td>$m = 12, \text{segments}$</td>
</tr>
<tr>
<td>$\theta = 0.9, \pi, \text{rad}$</td>
<td>$\eta_c = 0.8$</td>
<td>$\eta_t = 0.9$</td>
</tr>
</tbody>
</table>

$\omega_t = 0.32\, \pi\, \text{rad}$

8.6. Artificial pericardium

Besides providing compression directly to the heart, the RCCD has the versatility to incorporate other features into its structure to provide additional functionality. Inspired by the pericardium that surrounds the heart to protect and lubricate the heart’s movements [67], this section introduces a double-walled sac known as the artificial pericardium (APC) sandwiched between the RCCD and the heart (Fig. 8.5A). The APC consists of two thin layers made from a combination of silicone elastomer and non-stretchable fabric. This combination makes the layers soft, flexible and non-stretchable, allowing them to safely convey compression force from the RCCD to the heart. The cavity between the two layers is called the pericardial cavity and is filled with liquid. The APC contributes two more functions to the RCCD. First, it provides gentle contact with the epicardial surface of the heart and helps distribute compression evenly to eliminate local stress that may cause damage to the myocardium. Second, the APC pericardial cavity pressure can be utilised as a sensing feature to monitor compression force.

Figure 8.5B shows the manufacturing process to create an APC prototype. The inner and outer APC layers are semi-spheroidal shells made of Ecoflex 00-30 (Smooth-On, USA) and satin fabric using the molding technique. In detail, we made a semi-spheroidal shell from a piece of satin fabric and put it on a male mold and then dipped it in a female mold filled with uncured Ecoflex 00-30. The prototype specifications can be found in Table 8.1. Next, we assembled the two APC layers and sealed their bases together with Sil-Poxy (Smooth-On, USA) to create a sac. A fluid transmission tube was connected to the cavity of the fully assembled APC at its
apex to supply fluid and measure cavity pressure (note that a pressure sensor was connected to the fluid transmission tube). Finally, the APC was placed inside the RCCD so that the APC outer layer was in contact with the RCCD inner silicone layer. The integration was reinforced with sewing thread and Sil-Poxy.

**Figure 8.5.** Artificial pericardium (APC). (A) Design concept. APC is made of a silicone-fabric combination, filled with water and sandwiched between the robotic cardiac compression device (RCCD) and the heart. (B) Fabrication process. (C-E) Characteristics of APC show a relatively linear relationship between compression force and cavity pressure. (F) Ultrasound setup to examine contact between RCCD and porcine heart. (G) Ultrasound image of RCCD with APC showing continuous contact between APC and the heart. (H) Ultrasound image of RCCD without APC showing local contact of RCCD and the heart. RMSE, root mean square error; AMF, artificial muscle filament.
We investigate the sensing ability of the APC using the hoop stress setup shown in Fig. 8.3E. The RCCD with integrated APC was pressurised and worn on the hoop, then the input pressure was withdrawn using an 0.1 Hz sinusoidal signal with an amplitude that kept a minimum pressure at 0.1 MPa. The relationships of input pressure, cavity pressure and force are presented in Fig. 8.5C-E. The cavity pressure is inversely proportional to the input pressure with a fairly linear hysteresis profile. A hysteresis gap separates the pressurising and releasing curves (Fig. 8.5C). Similarly, the compression force has an inversely proportional relationship with the input pressure with a decent linear hysteresis profile. Interestingly, the hysteresis chart shows a very narrow gap between the two working phases (Fig. 8.5D). The RCCD was pressurised to 0.64 MPa and then reduced to 0.1 MPa to produce a cavity pressure of 6 kPa and a compression force of 13.1 N. Figure 8.5E shows the proportional relationship between the compression force and the cavity pressure. Although a hysteresis gap exists, the hysteresis profile is fairly linear as evidenced by a relatively small RMSE of 0.5 N between the linear fitting curve $F(P) = 2.735P - 3.271$ and the experimental data. The experimental results demonstrated the feasibility of using the APC as a sensing element to monitor the compression force of the RCCD on the heart.

We also built an experimental setup to examine the APC’s ability to provide soft contact and local stress relief to the epicardium. The setup consisted of a diagnostic ultrasound system (Mindray TE7, Shenzhen Mindray Bio-Medical Electronics, China) with an ultrasound probe (transducer) pointed towards the RCCD surrounding a fresh porcine heart (Fig. 8.5F). The porcine heart was coupled with a rotary fixture along the heart axis. The entire setup was submerged in a bucket of water. We examined the contact between the heart’s exterior and the RCCD with and without APC. The resulting ultrasound images are shown in Fig. 8.5G, H. The experimental results revealed that the APC provided continuous contact with the heart. Furthermore, the liquid-filled APC cavity acted as a damping layer to prevent the local stress generated by the compress-AMF of the RCCD (Fig. 8.5G). In contrast, local stress points were formed at the place where the compress-AMF made contact with the heart in the case of the RCCD without APC (Fig. 8.5H). The silicone layer of the RCCD may provide a little stress relief but it is too thin and soft to distribute the compression force evenly across the heart’s surface.
8.7. Customization for performance enhancement

The RCCD construction is highly customizable to provide the desired motion and force on the heart. For example, the two AMFs can be made longer to create a denser RCCD structure to increase compression. The predefined twist angle of the twist-AMF segments can be adjusted to manipulate the distribution of axial motion and twist angle change. This section investigates the enhancement of RCCD performance in terms of flow rate, stroke volume, compression force and output pressure when incorporating an additional AMF in its structure. The additional AMF is 755 mm long and has the same specifications as the two original AMFs (Table 8.1). It is helically arranged 4.5 revolutions around the silicone layer, beneath the twist-AMF similar to the original compress-AMF (Fig. 8.6A). The intersection points between the additional AMF and the twist-AMF are tightened with elastic strings. The inlets of the three AMFs are interconnected to receive a single source of hydraulic pressure. The modified RCCD is called

![Figure 8.6](image)

**Figure 8.6.** Customization of robotic cardiac compression device (RCCD) for performance enhancement. (A) Configuration A is the original RCCD and configuration B is the configuration A with an additional artificial muscle filament (AMF). (B-G) Performance comparison between two RCCD configurations in terms of flow rate, output pressure, stroke volume, average flow rate, maximum compression force and maximum blocked output pressure.
configuration B to distinguish it from the original RCCD, which is configuration A. The design concept and prototypes of the two RCCD configurations are shown in Fig. 8.6A. Configuration B has a denser structure than configuration A.

**Table 8.4. Performance comparison between two RCCD configurations**

<table>
<thead>
<tr>
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<tbody>
<tr>
<td>Maximum instantaneous flow rate (L/min)</td>
<td>2.8</td>
<td>3.5</td>
<td>25%</td>
</tr>
<tr>
<td>Maximum output pressure (mmHg)</td>
<td>50</td>
<td>58.9</td>
<td>17.8%</td>
</tr>
<tr>
<td>Stroke volume (mL) / average flow rate (L/min)</td>
<td>6</td>
<td>107.6/0.65</td>
<td>132.5/0.8</td>
</tr>
<tr>
<td>Speed (bpm)</td>
<td>12</td>
<td>82/0.98</td>
<td>98.8/1.19</td>
</tr>
<tr>
<td></td>
<td>15</td>
<td>70/1.05</td>
<td>81.3/1.22</td>
</tr>
<tr>
<td>Maximum compression force (N)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Loading pressure (MPa)</td>
<td>0.6</td>
<td>10.5</td>
<td>15.2</td>
</tr>
<tr>
<td></td>
<td>0.8</td>
<td>18.7</td>
<td>32.4</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>25.3</td>
<td>41.4</td>
</tr>
<tr>
<td>Maximum blocked output pressure (mmHg)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Loading pressure (MPa)</td>
<td>0.6</td>
<td>59.7</td>
<td>71.1</td>
</tr>
<tr>
<td></td>
<td>0.8</td>
<td>62.6</td>
<td>76.4</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>69.2</td>
<td>83.4</td>
</tr>
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</table>

The RCCD configuration B has undergone a similar series of experiments that we have performed with configuration A. The experimental results showing the performance of the two configurations are aggregated and presented in Fig 8.6B-G and Table 8.4. With sinusoidal signals that produce a maximum input pressure of 1.2 MPa, the RCCD configuration B generates higher instantaneous flow rates (25% enhancement) and output pressure (17.8% enhancement) than configuration A. The enhancement in stroke volume and average flow rate depend on the actuation speed with the highest increase of 23.1% at the lowest speed of 6 bpm. At the rates of 12 bpm and 15 bpm, the increases are 20.5% and 16.1%, respectively. The
compression force has the largest enhancement, reaching 73.3% at a loading pressure of 0.8 MPa. This loading pressure also enhances the blocked output pressure by 22%. With a loading pressure of 0.6 and 1 MPa, the compression force increases by 44.8% and 63.6% while the blocked output pressure increases by 19.1% and 20.5%, respectively. The experimental results clearly demonstrate the performance enhancement of the RCCD configuration B compared to configuration A.

8.8. Performance demonstration on porcine heart

Ideally, the RCCD should be tested on the hearts of live animals that have been chronically induced with HF because it is designed to support a failing (still working) and hypertrophic heart instead of a dead one. However, ex-vivo experiments on a dead heart still provide important insight into the heart-device interactions and the overall functions and performance of the device. Such preliminary results are crucial for device improvement and preclinical trials in the future. We implemented several variants of the RCCD on a deceased porcine heart to examine their performance in generating motion, flow rate and output pressure.

We supplied positive hydraulic pressure to expand the original RCCD (configuration A) and wore it on a porcine heart (Fig. 8.7A). The RCCD apex was in contact with the heart apex and the RCCD enclosed the entire epicardial surface of the heart from apex to base. The RCCD silicone layer was directly contacted with the epicardium. We adjusted the input pressure of the RCCD to achieve a firm wrap on the heart. The left and right ventricular chambers of the porcine heart were equipped with elastic balloons connected to a fluid tube, followed by a flowmeter and a water reservoir. A pressure sensor was attached to the fluid tube to measure output pressure. We then withdrew the input pressure using a 0.2 Hz (12 bpm) sinusoidal signal with an amplitude that kept a minimum input pressure at 0.1 MPa. The maximum input pressure was 1.2 MPa. The relationship between instantaneous flow rate and input pressure is shown in Fig. 8.7B, which is similar to the profile when the RCCD pumped an artificial univentricle (Fig. 8.3C). However, the flow rate magnitude is smaller in the case of pumping the porcine heart, which is 2 L/min pump out and 1.5 L/min fill in. These values are the maximum instantaneous flow rates. At the rate of 12 bpm, the RCCD coupled with the porcine heart produced a stroke volume of 61 mL, equivalent to an average flow rate of 0.73 L/min. It also generated an output pressure of 39.5 mmHg when the input pressure was completely withdrawn (Fig. 8.7C). A large hysteresis gap between the pump-out and fill-in phases can be seen in the
input-output pressure hysteresis chart. We observed that the RCCD provided the porcine heart with spatial motion including radial, axial and torsional simultaneously during both compression and expansion.

**Figure 8.7.** Performance demonstration of the robotic cardiac compression device (RCCD) on a porcine heart. (A) RCCD configuration A surrounds the heart to provide compression that produces flow rate (B) and output pressure (C). (D) The heart is wrapped with RCCD configuration A with an integrated artificial pericardium. (E) RCCD configuration B compresses the heart to produce stroke volume. (F) The heart is wrapped with RCCD configuration B.

We also implemented the original RCCD with integrated APC on the porcine heart (Fig. 8.7D). One obvious improvement is that the APC acted as a protective layer so the AMFs did not compress directly into the epicardium. As a result, the RCCD with APC experienced a smaller slip (displacement < 2 mm) on the cardiac surface compared with the RCCD without APC (14
mm displacement) after performing 15 cycles of compression and dilation. In addition, the APC cavity pressure offered another means of monitoring compression besides the input pressure. When the input pressure is completely withdrawn, the APC cavity pressure reached 7.8 kPa. Although the RCCD exerted spatial motion, we were unable to observe the heart movements due to the opaque APC. Furthermore, the RCCD with APC underperformed RCCD without APC, evidenced by an average flow rate reduction of about 18% and a decrease in output pressure by 23%.

The last RCCD variant that we implemented on the porcine heart was the modified RCCD (configuration B). Similar to configuration A, the RCCD configuration B was pressurised and worn on the porcine heart to induce motion. Figure 8.7E shows two working phases of the RCCD configuration B coupled with the heart, where stroke volume is indicated by the level difference of the red-dyed water column. A closer image of the contact between the device and the heart is shown in Fig. 8.7F. There is a thin silicone layer between the AMFs and the epicardium so the motion generated by the RCCD is well transferred to the heart. With an actuation speed of 12 bpm, the RCCD configuration B helped the heart produce an average flow rate of 0.83 L/min and an output pressure of 43.6 mmHg, which means an increase of 13.7% and 10.4% respectively compared to configuration A. These experimental results consolidate the previous conclusion that configuration B outperforms configuration A.

8.9. Discussion and conclusion

Driven by the large clinical need and limitations of current cardiac assist devices, this study introduced a new DCCD called robotic cardiac compression device to assist in the treatment of HF. The proposed RCCD is designed to surround the epicardium to actively provide compression to pump blood out of the ventricles during systole. It then expands during diastole, allowing blood to fill the ventricles. Inspired by the structure of cardiac muscles, especially the myocardial sheet and fibre arrangement from endocardium to epicardium, the RCCD has a semi-spheroidal shape made from a special arrangement of two artificial muscle filaments (AMFs), including a spirally arranged AMF to induce radial motion and an axially arranged AMF to induce axial and twisting movements. This design allows the RCCD to produce spatial motion including radial, axial and torsional movements, mimicking the natural motion of the heart. Previous researchers have highlighted that a cardiac assist device that conforms to the natural mechanics of the heart motion would be preferable for the treatment of HF [56, 68, 69].
The RCCD expanded by 34.5% radially, 14.5% axially and twisted by 32.2° when receiving positive hydraulic pressure. Subsequently, it contracted and twisted in the opposite direction when the input pressure was withdrawn. The twist angle achieved by the RCCD is larger than the maximum systolic torsion angle of the normal heart, around $8 \pm 2.1°$ [70]. That would not be a problem since the resulting twist angle can be adjusted by varying the predetermined twist angle of the AMF segments during fabrication. Also, the dynamic interaction between the device and the heart helps to reduce the twist angle. The relationships between output movements and input pressure are nonlinear with relatively large gaps in their hysteresis profiles (Fig. 8.2E-G). Nonlinear hysteresis is an inherent characteristic of AMFs constructed from silicone elastomers. The hysteresis gaps represent energy loss when the AMFs change their working phase from pressurising (accumulating elastic energy) to releasing (discharging the stored energy). Interestingly, no backlash exists in the hysteresis profiles, which means that the RCCD movements are highly responsive to hydraulic pressure.

The RCCD could produce a stroke volume from 70 to 107.6 mL depending on the actuation speed, which is comparable to the soft robotic sleeve made by Roche et al. with 84 mL [56]. However, the soft robotic sleeve could perform at 80 bpm owing to multiple PAMs while the RCCD achieved only 15 bpm. The use of long AMFs has limited the actuation speed. In vitro experiments revealed that the RCCD could produce a maximum flow rate of 2.8 L/min, an output pressure of 50 mmHg, a compression force of 25.3 N and a blocked output pressure of 69.2 mmHg. These results suggest that the RCCD underperformed compared to a normal heart but is comparable with a failing heart [60]. Therefore, the RCCD is intended to assist in the treatment of HF rather than taking over the heart function.

An empirical model based on experimental calibration has been developed to describe the 3D motion of the RCCD. Based on the input-output relationships established in the model, the resulting radial, axial and torsional movements can be fine-tuned by adjusting input parameters during the design and fabrication, such as the AMF lengths, the number of twisting segments and the predetermined twist angle. The model performed well in capturing the tendency of radial expansion, axial expansion and twist angle change with acceptable RMSEs. However, it was unable to closely follow every local change in the experimental data.

Inspired by the natural pericardium, we created a silicone artificial pericardium located inside the RCCD to provide soft and conformable contact with the heart. The double-layer structure with a fluid-filled cavity helped distribute compression from the RCCD to the heart evenly.
without local stress, thus potentially preventing myocardial damage. The cavity pressure of the APC can be used to monitor compression force with a fairly linear relationship (Fig. 8.5E). A modified RCCD with an extra helically arranged AMF was created to enhance the performance of the original RCCD. The modified variant showed significant improvement in all aspects including flow rate, stroke volume, output pressure and compression force (Fig. 8.6 and Table 8.4). Besides providing additional compression, the extra AMF also made the RCCD structure denser (reducing gaps between active filaments), thus conveying movements more effectively.

For demonstration, we implemented several RCCD configurations on a dead porcine heart. Overall, all RCCD configurations were able to wrap around the porcine heart from base to apex and induce 3D movements. Coupled with the heart, the original RCCD produced a stroke volume of 61 mL, a cardiac output of 0.73 l/min, and an output pressure of 39.5 mmHg. An enhancement from 10.4% to 13.7% was achieved by the modified RCCD. These results suggest that the performance of the RCCD is reduced when paired with the heart compared with an artificial univentricle. The underlying mechanism of these reductions includes the complex structure of the ventricular chambers that create air pockets when enclosing the elastic balloons. The compressible air pockets decreased compression efficacy. Another cause is that the RCCD slipped slightly on the heart surface during compression because the device was solely attached to the heart’s exterior by friction rather than adhered by glue or sewn by sutures.

The proposed RCCD has left some limitations for future improvement. Multiple shorter AMFs with a larger diameter and stronger contraction can be used to construct a more advanced RCCD with faster actuation speed and stronger compression than the current version. In addition, a denser structure of AMFs increases the transmission of motion from the device to the heart. Subsequent generations of the RCCD should have an effective binding method to maintain contact with the heart surface. Some potential adhesion approaches include the use of adhesive porous elastomers [49], suction cups [71] and biointegration of medical mesh [42]. A control system with integrated ECG is required to synchronise the device’s movements with cardiac motion [42, 56]. Subsequently, a preclinical trial in live animals should be conducted to evaluate the therapeutic efficacy of the device.

In conclusion, this study has introduced a proof-of-concept robotic cardiac compression device made of artificial muscle filaments and silicone elastomers to assist in the treatment of advanced HF. The proposed cardiac assist device surrounds the epicardium to actively provide compression during systole and expansion during diastole. The device could generate spatial...
motion including radial, axial and torsional movements to mimic the natural motion of the heart. An artificial pericardium is developed to provide gentle contact with the heart and for sensing purposes. The robotic cardiac compression device structure can be customised to enhance performance. The proposed device is implemented on a dead porcine heart to induce 3D motion, output flow rate and pressure. The design concept and preliminary results of this study are expected to inspire further improvement as well as extend the development of active cardiac assist devices.

References


Chapter 8. Robotic cardiac compression device


Chapter 8. Robotic cardiac compression device


58. M. R. Moreno et al., "Assessment of minimally invasive device that provides simultaneous adjustable cardiac support and active synchronous assist in an acute heart failure model," *Journal of Medical Devices*, vol. 5, no. 4, 2011.


Chapter 9

Discussion and conclusion

The field of soft robotics, with the impressive developments of soft artificial muscles (SAMs), smart textiles, wearable assistive devices, flexible surgical instruments, biomimetic soft robots and cardiac assist devices, has become an emerging research topic worldwide. Stemming from unmet needs in the healthcare system and limitations of current soft actuators, this thesis proposed to develop a new soft actuator called hydraulic filament artificial muscle (HFAM) for multiple robotic and medical applications. Many aspects of the HFAM including design, fabrication, characterisation, mathematical modeling and control strategy have been thoroughly presented throughout the thesis. In parallel, proof-of-concept devices made of HFAMs for a wide range of applications have been demonstrated. The following sections summarise the thesis’s achievements, limitations and suggestions for future development.

9.1. Thesis summary

The development of HFAM, its versatile configurations and devices for robotic and medical applications have been presented throughout the thesis.

Chapter 1 provided an overview of the research topic, which is a new type of SAM called HFAM for multiple applications. The clinical need for a flexible, efficient and versatile actuation mechanism as well as the limitations of current artificial muscle technology have driven the development of HFAM. This chapter also outlined the research aims and structure of the thesis.

Chapter 2 presented a literature review of the current status of SAM technology in four main categories based on their excitation sources, including electrically-driven, magnetically-driven, thermally-driven and pressure-driven artificial muscles. The advantages and limitations of SAMs in each category have been discussed. This chapter then provided an overview of four areas of interest for HFAM implementation. These include robot-assisted devices for
rehabilitation, wound closure technology, shape-morphing soft robots and smart textiles, and cardiac assist devices for the treatment of heart failure.

Chapter 3 introduced the fundamental development of HFAM including design concept, working principle, manufacturing methods, characteristics, mathematical models and demonstration. The HFAM is a thin, scalable and flexible filament made by inserting a silicone tube inside an extension coil spring. It expands longitudinally and accumulates elastic energy when receiving positive hydraulic pressure. It then contracts and discharges elastic energy when the input pressure is released. The HFAM achieved high performance such as 0.15 MPa stress, 246.8% strain and 62.7% energy efficiency. It was also highly durable and responsive. A mathematical model was developed to describe the proportional relationship between elongation and force of the HFAM. The model showed good performance by closely following the experimental data to 100% strain. The HFAM was experimentally demonstrated to have small energy losses regardless of the tortuous transmission paths, which is advantageous compared with tendon-pulley and tendon-sheath mechanisms. This chapter also presented the first application of the HFAM, which was a wearable soft robotic glove that used the HFAM as the actuation mechanism. The robotic glove assisted in grasping and holding objects of various shapes and sizes.

Chapter 4 introduced the second application of the HFAM – smart surgical sutures for wound closure. Unlike conventional passive surgical sutures, smart surgical sutures are active filaments that can contract to tighten tissues once the target suturing procedure is finished. The HFAM is pressurised and blocked at a certain elongation and then is used in suturing procedures. Subsequently, by releasing the input pressure, the HFAM is shortened, which generates compression to tighten the detached tissue. Depending on the target procedures, the original length and initial elongation of smart surgical sutures can be determined during fabrication. Three smart surgical suture prototypes with the same outer diameter (OD) of 1.49 mm and different lengths achieved an elongation of 107%. A modified mathematical model that accounts for the dynamic Young’s modulus of the silicone tube was developed to describe the elongation-force relationship of the HFAM. The model performed well in capturing the experimental data up to 200% strain, much better than the model in Chapter 3. The proposed smart surgical sutures can be knotted like conventional surgical sutures or used with the help of anchors at two ends to secure tissues. Three different types of anchors have been created and tested for puncturing and holding force on porcine stomach tissue. The smart surgical sutures
in two different sizes (OD 1.49 and 0.8 mm) have been demonstrated ex-vivo for perforation closure and tissue folding (a weight loss procedure) on porcine stomach and colon.

Chapter 5 studied the twisted and braided configurations of HFAMs to enhance their performance in terms of elongation and force. When receiving input pressure, these configurations reduce the twist angle, which contributes to an increase in the overall longitudinal extension. The elongation enhancement improves the contraction force exertion. The experimental results showed the twisted and braided configurations provided an elongation enhancement of 10.4% and 14.6% respectively, compared with HFAMs in a straight arrangement. As a result, the contraction force of these twisted and braided variants increased by 11.2% – 16.1%. Mathematical models to describe the elongation-force relationship of twisted and braided configurations based on the fundamental model in Chapter 3 and the geometric arrangements of these variants have been proposed and validated. A pair of HFAMs was deployed to manipulate the full range of motion of a 3D-printed index finger. Similarly, another pair of twisted HFAMs was used to mimic the bicep and tricep muscles to drive the elbow joint model over a 120° range of motion. This chapter also introduced a tubular structure made of eight HFAMs by the hollow round braiding technique. The tubular structure simultaneously expands in radial and axial directions upon pressurisation. It was implemented for multiple applications including a compression sleeve to provide massage therapy, a hollow cylindrical support structure (potential for stents or gastrointestinal support devices for endoscopic surgery) and a tubular gripper to retrieve objects in a confined space (potential for foreign object retrieval in the human gastrointestinal tract).

Chapter 6 presented a new type of smart textile made of thin, long HFAMs using knitting, weaving and sticking techniques. Smart textiles inherently have the flexibility and breathability of conventional textiles while providing the ability to control their shape and deformation. Several prototypes were created and the area expansion was experimentally measured when pressurised, including a knitted sheet with 35%, a bidirectional woven sheet with 108%, a unidirectional woven sheet with 65% and a circular woven sheet with 25% area expansion. Mathematical models to describe the relationship between elongation and force of knitted and woven sheets have been developed and experimentally validated. The planar smart textiles can be mechanically programmed to induce multimodal motion and shape-shifting capability such as compression sleeves, s-shapes, hyperbolic paraboloids, hollow structures, bidirectional bending, as well as bending and expanding independently. This chapter also introduced the concept of sticking HFAMs to fabrics to reconfigure them from passive to active and
controllable. The HFAMs are strategically stuck and mechanically programmed onto a piece of conventional fabric to transform its structure from a 2D plane to a 3D structure. This simple and easy method has been used to provide several biomimetic soft robots including a four-legged structure, a butterfly and a flower. This approach enables the development of deployable structures, bio-inspired soft robots and wearable assistive technologies.

Chapter 7 investigated the modeling, control and application aspects of HFAM-based smart textiles. A woven smart textile was created and experimentally characterised. The woven smart textile exhibited nonlinear hysteresis behavior expressed in the elongation-volume relationship. The hysteresis gap was widened with increased input amplitude and frequency. A simple nonlinear hysteresis model based on the modified Bouc-Wen model was developed. The proposed model was then implemented into a feedforward control scheme based on inverse compensation to capture the output elongation of the woven smart textile. In addition, an adaptive controller was developed to improve output tracking performance. Both the feedforward and adaptive controllers showed remarkable performance in tracking output elongation, as evidenced by a reduction in tracking error of up to 90% compared to the uncompensated system.

This chapter also introduced a portable smart compression device to provide compression therapy, potentially used for the treatment of venous disease or compromised peripheral blood flow. The proposed compression device featured a compact actuation unit that drives a low-profile and easy-to-wear smart compression sleeve to provide compression to the human extremities. The device was controlled wirelessly with negligible latency via Bluetooth using a user-friendly smartphone application. The actuation unit weighs 480 g and can be worn comfortably around the waist. The smart compression sleeve could provide 19.5 N compression force and took 5 s to complete a cycle of compression and decompression with low power consumption. Besides providing active compression, the proposed woven smart textile’s functionality can be expanded by integrating functional features such as soft contact force sensors and other electronic components.

Chapter 8 introduced a robotic cardiac compression device made from a special configuration of HFAMs to assist in the treatment of heart failure. The device was designed to wrap around the heart’s surface to provide active compression to assist blood ejection from the ventricles. The device was equipped with two HFAMs arranged in circumferential and axial directions to generate spatial motion (radial, axial and torsional movements) that mimicked the natural
motion of the heart. When positive hydraulic pressure was applied, the robotic cardiac compression device expanded by 34.5% radially, 14.5% axially and twisted by 32.2°. The device was then contracted and twisted in the opposite direction to return to its original state when the input pressure was withdrawn. Related to performance, the device achieved an actuation speed of 15 beats per minute, provided a stroke volume of 70 – 107.6 mL, a maximum flow rate of 2.8 L/min, an output pressure of 50 mmHg, a compression force of 25.3 N and a blocked output pressure of 69.2 mmHg.

An empirical model based on experimental calibration to describe the spatial motion of the device was developed. The model showed good performance in following the experimental data with relatively small errors. A silicone artificial pericardium was created to provide soft contact and distribute compression from the device to the heart surface, helping to reduce local stress on the epicardium. The cavity pressure of the artificial pericardium was proportional to the compression force with a fairly linear relationship, which greatly facilitated its use as a sensing component. This chapter also introduced a modified robotic cardiac compression device with an additional HFAM arranged in the circumferential direction. The modified device outperformed the original device in many aspects. Several configurations of the proposed device have been implemented to surround a dead porcine heart to provide compression and 3D motion.

Both modes of operation (extension and contraction) of the HFAM have been utilised throughout the proposed devices. The contraction force of HFAM has been implemented in most devices that generate strong mechanical work such as wearable robotic glove, smart surgical sutures, tubular gripper, compression garment, and robotic cardiac compression device. In the case of smart textiles, shape-morphing and fabric reconfiguration structures, the extension force became dominant to induce movement while the contraction force restored their original shapes.

Finally, chapter 9 summarises the achieved results, limitations and future directions of the thesis.

9.2. Limitations and future directions

Overall, this thesis has investigated the development of a new type of artificial muscle and its versatility for a wide range of robotic and medical applications. Therefore, the primary focus
of this thesis was to study many aspects of the proposed artificial muscle and demonstrate the feasibility of its use for broad practical applications rather than focusing on a comprehensive study of a particular medical application. This approach led to the proposal of many proof-of-concept devices for applications but with shallow development. The following segments discuss limitations and recommendations for future in-depth development of applications based on the proposed artificial muscles.

The first limitation belongs to the HFAM structure. Stainless steel coil springs provide strong contraction force but require high input pressure to operate. For example, an HFAM with an outer diameter of 1.49 mm needed 1.3 MPa to generate 80% elongation. To reduce the input pressure, more research on the coil material is required. Some potential materials include thermoplastic polymers, nitinol and carbon fibre composites. Another approach to reducing the coil spring constant is to use coils with smaller wire diameters. In addition, different coil dimensions and configurations should be explored to further optimise the HFAM stiffness and performance. Future development will provide a guideline on material selection and specification (both silicone tubing and coil springs) for specific requirements.

The proposed wearable soft robotic glove was demonstrated with just one user for proof-of-concept purposes. A large user study is needed to evaluate the usability, effectiveness and any drawbacks of the robotic glove. Such clinical trials require participants with impaired hand movement to grasp and hold different objects with increasing weight to assess successful rates, comfort, actuation speed and ease of manipulation. A trial protocol, recruitment plan and user data management plan are required prior to an ethics application. Furthermore, the current version of the robotic glove can actively assist three fingers (thumb, index finger and middle finger). Future development should consider supporting the whole hand.

The proposed smart surgical suture has a relatively large diameter (0.8 mm) and requires high force to puncture the tissue. Therefore, reducing the suture size is crucial before preclinical trials. The distance between the two anchors of the smart surgical suture was fixed, making it difficult to provide the desired tightening force to the tissue. A simple mechanism to dynamically adjust the distance between anchors is required to control the suture tension. An investigation into biocompatible materials to produce smart surgical sutures or to provide a coating layer is essential for preclinical and clinical trials in the later stages of development. Further directions also include in-vivo experiments on live animals to evaluate the performance and safety of the smart surgical sutures.
Chapter 9. Discussion and conclusion

The use of twisted configurations of the HFAM to manipulate finger and elbow joints of 3D-printed models has been demonstrated in Chapter 5. Although the proposed HFAM variants could produce the full range of motion of the models, they were unable to provide position control for any desired bending angles. Future development should incorporate a sensing feature (e.g. liquid metal microtubule sensor) into the HFAM structure for position control. Another approach is to use the input pressure to monitor the output elongation, assuming that a hysteresis model is available. The tubular structure created by the braiding technique has been demonstrated for various applications. However, an in-depth study is required to evaluate the safety and effectiveness of the device including preclinical trials of the tubular gripper to retrieve foreign objects inside the gastrointestinal tract or an in-vivo study of the tubular structure to support hollow organs during endoscopic surgery.

Although the smart textile prototypes in Chapter 6 were manually created, it is possible to make smart textiles from thin, long HFAMs by machine. A study implementing HFAMs into sewing, knitting or embroidery machines to create smart textiles is strongly recommended. Machine-made products will have reliable quality and reproducibility, shorter manufacturing times and larger sizes. The multimodel deformation and shape-shifting ability of smart textiles have been demonstrated. However, they were not implemented into a specific application. Therefore, a comprehensive study of the shape-shifting approach for a particular application is required in future work. Furthermore, it is desirable to develop soft sensing features for smart textiles and shape-morphing structures to have better control over their deformation. Potential sensing components include liquid metal microtubule sensors, serpentine capacitive sensors, or utilise input pressure as a sensing feature.

The limited actuation speed of the current experimental setup hindered the tracking performance of the proposed feedforward and adaptive controllers. A faster motorised linear slider is required to exploit the full potential of the controllers. It is recommended that future work should combine both feedforward and adaptive controllers to further reduce output tracking errors for high precision control requirements. Related to the portable smart compression device, a faster and stronger motor is required to increase the compression speed. It is also preferable for the compression sleeve to have a stronger compression force by using a bigger or stiffer HFAM. Future development of the portable smart compression device should consider incorporating feedforward and adaptive control strategies into the control system to improve output tracking and precision control requirements. Furthermore, patients may benefit from the peristaltic motion of the compression sleeve with multiple HFAMs and smart control.
strategies. A user study is recommended to evaluate the usability, effectiveness, and potential disadvantages of the portable smart compression device.

For an advanced robotic cardiac compression device with faster actuation speed and stronger compression, future work should use shorter but more powerful HFAMs to construct the device and invest in a faster motorised linear slider to drive the device. An effective adhesion method should also be investigated to establish contact between the device and the cardiac epicardium. As a result, the device can actively assist the heart during diastole to fill the ventricles with blood. A control system with an integrated electrocardiogram is required to synchronise the device’s movements with the heart. It is highly recommended to incorporate feedforward and adaptive control strategies into the control system to improve precision control requirements. In addition, in-vivo experiments on live animals should be conducted in the future to evaluate the safety and performance of the proposed robotic cardiac compression device.

Although open-loop control is sufficient to demonstrate the feasibility of proof-of-concept devices (wearable robotic glove, smart compression garment, robotic cardiac compression device) in this thesis, closed-loop control with feedback will be more beneficial for practical applications. Therefore, it is highly recommended to investigate closed-loop control strategies for a particular application. For example, the wearable robotic glove can be equipped with contact force sensors so that the control system can regulate the grasping force applied to objects. Similarly, force or pressure sensors can be integrated into the compression sleeve to sense and control the compression force applied to the patient’s extremities to improve treatment efficiency and safety. The artificial pericardial cavity pressure of the robotic cardiac compression device can be used as a feedback signal to control the compression force of the device on the heart.

All the proposed devices in this thesis are at the proof-of-concept stage aiming to demonstrate the feasibility of the design concept and working principle. Therefore, future development of a particular device based on HFAM should include an extensive durability test to demonstrate the long-term durability of the device under actual working conditions. The durability test will provide insights into the device structure, facilitating improvement and optimisation.
9.3. Conclusion

This thesis has introduced the development of hydraulic filament artificial muscles for robotic and medical applications. The new artificial muscle expands longitudinally when receiving positive hydraulic pressure and contracts when the input pressure is withdrawn. It is flexible, has a high length-to-diameter ratio, high elongation and high energy efficiency. The proposed artificial muscle has been demonstrated in many applications, including a wearable soft robotic glove to assist in grasping various objects, smart surgical sutures for wound closure and tissue plication, soft actuators for joint manipulation, a soft tubular gripper to retrieve objects in confined spaces, shape-shifting smart textiles for biomimetic soft robots, smart compression devices to provide massage therapy and robotic cardiac compression devices to assist in the treatment of heart failure. These proof-of-concept devices contribute to expanding knowledge and are expected to inspire further development of medical robotics in the future.