New types of McKibben artificial muscles

Danial Sangian

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New Types of McKibben Artificial Muscles

A thesis submitted in fulfillment of
the requirements for the award of the degree

Doctor of Philosophy

from

University of Wollongong

by

Danial Sangian


School of Mechanical, Materials and Mechatronic Engineering
Intelligent Polymer Research Institute
Wollongong, Australia
September 2016
Declaration

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

Danial Sangian
September, 2016
Acknowledgements

I would like to thank my friend and principal supervisor Professor Geoffrey Spinks for his encouragement, support and difficult questions. I would also like to thank my co-supervisor Dr. Sina Naficy for his assistance in experimental and modeling work, as well as his friendly support. My gratitude goes to the faculty of engineering of the University of Wollongong for providing me with a scholarship to cover my expenses.

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I would like to express my sincere gratitude to my parents, Sima and Hossien and my sisters Dayana and Diba for their patience and emotional support.
Abstract

Actuators are devices that exhibit reversible change in their shape or volume or generate force when externally stimulated. Because of their very similar operation style to biological muscles, actuator materials are also known as artificial muscles. These materials are in demand for many applications, such as medical devices and robotics. These applications normally require an inexpensive actuator system that can offer high force, high strain, and high power density in a relatively short period of time. The device packaging and size of the actuator are also important parameters as currently most of the applications desire very compact and lightweight systems. Furthermore, low electricity consumption also as a last requirement has a significant effect on the actuation system by increasing the efficiency of the entire system. Producing all of the above requirements in one device is currently a challenge for engineers and scientists.

In this thesis, a new contractile artificial muscle system is introduced than can offer most of the above requirements to satisfy the current expectations of these devices. Chapter 1 of this thesis focuses on a literature review of prominent available artificial muscles and comparing them with biological muscle performance for better understanding of their advantages and disadvantages. Chapter 2 investigates the effect of the inner tube material and muscle geometry on a small hydraulic McKibben artificial muscle as well as the possibility of running this system with a compact, low voltage water pump. This chapter also introduces a new equation that is able to predict static muscle performance notably more accurately than previous models. Chapter 3 illuminates the possibility of three-dimensional printing the braided sleeve used in McKibben artificial muscles to have more control on the manufacturing process of such devices. In Chapter 4, the fluid normally used in conventional McKibben muscles is substituted with a temperature sensitive material to eliminate the need
of the pump/compressor and piping to introduce a more compact device. The new muscles were stimulated either by immersing in a hot water bath or using a heating filament. A contraction strain of 9% and 2 N isometric force were produced. A new equation is also introduced to predict the performance of this type of McKibben muscles with temperature as the driving force. Chapter 5 introduces a novel miniature type of McKibben artificial muscle by using a conductive braided sleeve and eliminating the need for the inner tube. The electricity consumption of this muscle is as low as 2.5 V. The muscle weight is only 0.14 gr with a diameter of 1.4 mm. The muscle generates a tensile stress of 50 kPa and contraction strain of 10%. Finally, Chapter 6 concludes this study and also represents some potential future works.
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<td>Electroactive polymers</td>
</tr>
<tr>
<td>SMAs</td>
<td>Shape memory alloys</td>
</tr>
<tr>
<td>CPS</td>
<td>Conducting polymers</td>
</tr>
<tr>
<td>CNT</td>
<td>Carbon nanotube</td>
</tr>
<tr>
<td>LCE</td>
<td>Liquid crystal elastomer</td>
</tr>
<tr>
<td>SWNTs</td>
<td>Single walled CNTs</td>
</tr>
<tr>
<td>MWNTS</td>
<td>Multiwalled CNTs</td>
</tr>
<tr>
<td>TBA.PF$_6$</td>
<td>Tetraethylammonium hexafluorophosphate</td>
</tr>
<tr>
<td>DEAs</td>
<td>Dielectric elastomers</td>
</tr>
<tr>
<td>$S_z$</td>
<td>Strain of DEAs</td>
</tr>
<tr>
<td>$Y$</td>
<td>Young modulus</td>
</tr>
<tr>
<td>$\varepsilon_r$</td>
<td>Relative permittivity</td>
</tr>
<tr>
<td>$\varepsilon_0$</td>
<td>Vacuum permittivity</td>
</tr>
<tr>
<td>$Z$</td>
<td>Thickness of membrane</td>
</tr>
<tr>
<td>PVDF</td>
<td>Poly vinylidene fluoride</td>
</tr>
<tr>
<td>NiTiNoL</td>
<td>Nickel-titanium alloy</td>
</tr>
<tr>
<td>PAMs</td>
<td>Pneumatic artificial muscles</td>
</tr>
<tr>
<td>HAMs</td>
<td>Hydraulic artificial muscles</td>
</tr>
<tr>
<td>LDPE</td>
<td>Low density polyethylene</td>
</tr>
<tr>
<td>$F$</td>
<td>Force</td>
</tr>
<tr>
<td>$P$</td>
<td>Pressure</td>
</tr>
<tr>
<td>$\varepsilon$</td>
<td>Contraction strain</td>
</tr>
<tr>
<td>$R_0$</td>
<td>Initial radius</td>
</tr>
<tr>
<td>$\alpha$</td>
<td>Braid initial angle</td>
</tr>
<tr>
<td>$K_F$</td>
<td>Force fitting parameter</td>
</tr>
<tr>
<td>$K_\varepsilon$</td>
<td>Contraction strain fitting parameter</td>
</tr>
<tr>
<td>PPS</td>
<td>Polyphenylene sulfide</td>
</tr>
<tr>
<td>Symbol</td>
<td>Description</td>
</tr>
<tr>
<td>--------</td>
<td>-------------</td>
</tr>
<tr>
<td>PVC</td>
<td>Polyvinyl chloride</td>
</tr>
<tr>
<td>Eq</td>
<td>Equation</td>
</tr>
<tr>
<td>$P_{th}$</td>
<td>Threshold pressure</td>
</tr>
<tr>
<td>$P_{el}$</td>
<td>Elastic pressure</td>
</tr>
<tr>
<td>$r$</td>
<td>Inner radius of the bladder</td>
</tr>
<tr>
<td>PCL</td>
<td>Polycaprolactam</td>
</tr>
<tr>
<td>M</td>
<td>Muscle</td>
</tr>
<tr>
<td>T</td>
<td>Temperature</td>
</tr>
<tr>
<td>$\alpha$</td>
<td>Thermal expansion paraffin</td>
</tr>
<tr>
<td>$\gamma$</td>
<td>Thermal pressure coefficient</td>
</tr>
<tr>
<td>DC</td>
<td>Direct current</td>
</tr>
<tr>
<td>C</td>
<td>Cover factor</td>
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<tr>
<td>3D</td>
<td>Three dimensional</td>
</tr>
<tr>
<td>$\Theta_0$</td>
<td>Initial braid angle</td>
</tr>
<tr>
<td>$W_y$</td>
<td>Yarn width</td>
</tr>
<tr>
<td>$\sigma$</td>
<td>Radius of the pores</td>
</tr>
<tr>
<td>N</td>
<td>Number of the threads</td>
</tr>
<tr>
<td>$\Theta$</td>
<td>Surface tension</td>
</tr>
<tr>
<td>V</td>
<td>Volume</td>
</tr>
<tr>
<td>$k$</td>
<td>Coefficient of compressibility</td>
</tr>
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</table>
Chapter ONE

Introduction to artificial muscles
1.1. Research background

The concept of creating artificial muscles [1-3] mainly comes from two important applications. Firstly, there is a need for assisting human movement in cases of injury or disability [4-6]. Secondly, artificial muscles could be useful for specific applications necessitating a human-like softness, such as miniaturized soft tools for small surgeries and soft arms for working in cooperation with people[7, 8]. The first stage in creation and development of an artificial muscle system is to recognize the principal engineering properties of biological muscles. The main engineering function of this natural actuator [9, 10] is to generate useful force and displacement by converting chemical energy into mechanical energy in a relatively short period of time. Moreover, this naturally developed machine is robust, lightweight[11], and exhibits an efficient delivery system to supply glucose and oxygen as combustion and withdraw the heat and waste. Biological muscles (natural muscles) are also significantly efficient, fast, self-repairable as a result of millions of years of biological evolution[12]. Biological muscles normally offer three different types of actuation movements in nature: torsional flagellum as an oldest (3500 million years ago) actuator enables some bacteria to propel themselves in liquid environment [13, 14], contractile leg muscles assist kangaroos to jump [14] and bending tail helps fish to swim [15]. Tensile contractile (linear) is the most common movement in biological muscle of humans and animals, allowing complex and agile movements as in jumping and lifting [16]. Linear mammalian skeletal muscles often display very unique properties (Table 1.1). These properties have never been completely mimicked by any manmade artificial muscle technology to date.
Table 1.1. Engineering properties of contractile (linear) biological muscles [12].

<table>
<thead>
<tr>
<th>Properties</th>
<th>Typical value</th>
<th>Maximum value</th>
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<tbody>
<tr>
<td>Tensile strain (%)</td>
<td>20-40</td>
<td>&gt; 40</td>
</tr>
<tr>
<td>Tension intensity (kPa)</td>
<td>100 (sustainable)</td>
<td>350</td>
</tr>
<tr>
<td>Work density (kJ/m$^3$)</td>
<td>8</td>
<td>40</td>
</tr>
<tr>
<td>Density (kg/m$^3$)</td>
<td>1037</td>
<td></td>
</tr>
<tr>
<td>Strain rate (%/s)</td>
<td>50</td>
<td>50</td>
</tr>
<tr>
<td>Power to mass (W/kg)</td>
<td>50</td>
<td>200</td>
</tr>
<tr>
<td>Efficiency (%)</td>
<td>20-25</td>
<td>40</td>
</tr>
<tr>
<td>Cycle life</td>
<td></td>
<td>10$^7$</td>
</tr>
<tr>
<td>Modulus (MPa)</td>
<td>10-60</td>
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Fuel engines are generally most efficient for continuous operating systems but are not desirable where frequently interrupted movement is required, such as valves or walking robots. Traditionally, large stroke actuation is achieved with piston–cylinder actuators that offer high forces. These systems however are difficult to seal, heavy and bulky, especially considering the pumps and compressors needed and they also suffer from static friction [17]. Electric motors as another alternative are heavy in mass compared to biological muscles because of the carrying power supply (batteries), making them unfavorable for some medical, robotic, and fluidic applications [18]. Piezoceramic materials [19] also generate high power densities and are fast [20] but only generate very small strains around 0.1%, which is far behind of biological muscles and therefore unpractical for most areas. Polymers have attractive and interesting properties in comparison to inorganic materials, which make them a suitable candidate to replace or simulate biological muscles [21-23]. They are lightweight, inexpensive, flexible, sensitive to extra stimuli and easily manufactured. However, a wide range of challenges, such as slow response, low heat toleration, short cycle life, use of electrolytes and low energy efficiency remain with these materials.
As a consequence, designing and developing new types of artificial muscles [1, 24] that can mimic the skeletal muscles has attracted significant attention among engineers and scientists [25]. The task can be done by conducting further investigation in the development and improvement of the performance of existing artificial muscles or alternatively inventing new types of these materials.

In this thesis, artificial muscles have been divided into four main categories. The first group includes artificial muscles that respond to an electric field or ionic changes. These are commonly known as electroactive polymers (EAPs). The second group is shape memory alloys (SMAs) that are sensitive to temperature because of phase changes that occur in their crystalline structure. Third is a type of shape memory polymers, which are sensitive to extra stimuli and show reversible change in their shape. The fourth group normally operates with pressurized fluid and is known as fluidic actuators.

1.2. EAP

Electroactive polymers (EAPs) are type of materials, which exhibit a change in size or shape when stimulated by an electric field. Actuators and sensors are the most common applications of this type of materials. Table 2 divides electroactive polymers to two main groups according to the type of stimulation.
Table 1.2. List of leading EAP materials.

<table>
<thead>
<tr>
<th>Ionic EAP</th>
<th>Electronic EAP</th>
</tr>
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<tr>
<td>Conducting polymers (CP)</td>
<td>Dielectric EAP</td>
</tr>
<tr>
<td>Carbon nanotubes (CNT)</td>
<td>Liquid crystal elastomers (LCE)</td>
</tr>
<tr>
<td></td>
<td>Ferroelectric polymers</td>
</tr>
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</table>

1.2.1. Ionic EAP (Conductive polymers, CNT actuators)

In these particular materials generating actuation is due to mobility of ions within the polymer phase. Generally, swelling or contracting occurs when an applied field drives the ions and entrained solvent exchange between the polymer and an external electrolyte reservoir. In the case of electronically conducting polymers such as conducting polymers and CNTs there is strong local fields while overall voltage is low as ions serve to balance charge generated on these conductors once potential is applied. The operating voltages in these materials are low (1-5 V), however, as a result of narrow space between ions and electronic charges, and the large amount of charge that can be transferred, the energies are still high.

1.2.1.1. Conductive polymers

Conducting polymers normally show insulating behavior in the undoped state and semi-conducting when doped with donor or acceptor ions [26, 27]. Doping is normally achievable chemically or electrochemically. A wide range of applications such as polymer light-emitting diodes, drug delivery systems, energy storage, electrochemical sensors and actuation devices
are identified for conductive polymers [28-34]. Polypyrrole and polyaniline (Fig. 1.1) are two typical conducting polymers [35, 36]. These conducting polymer artificial muscles use the dimensional changes as a result of electrochemical ion insertion and de-insertion, possibly along with associated solvating species [37, 38].

![Chemical structure of two important conductive polymers (A) polypyrrole (B) polyaniline.](image)

Figure 1.1. Chemical structure of two important conductive polymers (A) polypyrrole (B) polyaniline.

Two electrodes are needed for these systems to complete the electrochemical cell. Both electrodes can include conducting polymers; both can be used as artificial muscles. The expansion process is mainly perpendicular to the polymer chain orientation as a result of ions and accompanying solvent locating between polymer chains (Fig. 1.2). [39-41]
Contractile strains obtained by conductive polymers such as polypyrrole, polyaniline and polythiophene are typically 2-10 %, however recent work by Kaneto and coworkers indicate that actuator strains can reach 40% [44-46]. Actuation rates are typically low (<1%/s) as a result of relatively slow migration of ions within the polymer and the large degree of doping [42, 47]. However, the actuation rate can be increased to 10 %/s by using metal contacts, porous polymers, fast charging methods or thin films and fibers while the strain rate of biological muscles is around 500 %/s [48-50]. Work densities of these materials normally [51] approach 100 MJ/m³. Operating voltages are ~2 V, higher voltages up to 10 V have been also used to increase the actuation rate. The significant advantage of conducting polymers over other available artificial muscles is their low operating voltage as well as their abilities to produce higher strains and lower cost than CNTs.

1.2.1.2. Carbon nanotubes (CNTs) actuators

In general, single-walled CNTs (SWNTs) are known as a single layer of graphite (graphene) rolled into a cylinder of nanometer diameter (Fig. 1.3A). Multiwalled CNTs (MWNTs) are
nested SWNTs (Fig. 1.3B). Individual SWNTs or very long MWNTs have dramatic mechanical properties [52, 53]. The tensile modulus of SWNTs (640 GPa) is close to diamond, while they exhibit tensile strength of 20-40 GPa, which is roughly ten times higher than any other kind of continuous fiber [18].

Figure 1.3. A graphene sheet rolled into a nanotube (A) and a MWNT (B).

The mechanical properties in the above range are reported only for individual SWNTs and the properties reduce in yarns and sheets shapes, which limit the performance of actuators based on nanotube yarns or sheets. Similar to conducting polymer actuators the dominate actuation mechanism is ion exchange between the porous CNT assembly and an external electrolyte due to charge injection into CNTs. A voltage is normally applied between an actuating nanotube electrode and a counter electrode, via an ion containing solution, where the counter electrode can be another CNT, leading to charging. Electrostatic repulsive forces between similar charges on the CNTs stimulate the nanotubes to lengthen and expand by
operating against the stiff carbon-carbon bonds in the nanotubes. Ions from the external electrolyte also migrate to the surface of the CNTs to form the electrical double layer. The adsorption of these ions can generate a swelling pressure within the pore space of the CNT electrode causing swelling of the electrode. The contractile strains of these specific materials are low (<2%) [39], as a result of the CNTs extreme stiffness.

Stresses up to 100 times more than biological muscles were achieved by these materials. By using thin films and fibers with porous nature these actuators can show low response times and effective strain rates of <10 ms and 19%/s, respectively. However, with increasing nanotube yarn or sheet thickness, the achievable response rate drops, raising interelectrode separation and decreasing electrolyte ionic conductivity. Power to mass ratios of 270 W/kg is achievable (half that of a high revving electric motor) [55]. Work densities in CNT fibers and yarns are ~1 MJ/m^3 (as in dielectric elastomers and ferroelectric polymers) [56]. High work density and good temperature stability (>450°C in air, >1000°C in an inert environment) are the unique properties of CNTs which make them a significant candidate for applications where weight and temperature are important, such as aerospace field [57, 58].

Lately, Foroughi and his co-workers [13] have introduced an electrolyte-filled twist-spun carbon nanotube yarn with a size thinner than a human hair. This unique torsional artificial muscle operates by electrochemical double-layer injection, offering a reversible 15,000° rotation and 590 revolutions per minute [13] with 0V to 5V applied voltage (0.2 M TBA.PF_6 in acetonitrile). The system simply contains a twisted MWNT yarn which is partially immersed in an electrolyte and a counter electrode (Fig. 1.4). Since the yarn is tethered at both ends, applying voltage between the electrodes forces the yarn to rotate and subsequently generates torsional actuation. However, this particular muscle suffers from low contractile strain around 3.4 % which is significantly behind of biological muscles. Therefore, this system is unpractical, where linear muscle like behavior is required.
Recently, in order to eliminate the need of counter electrode and electrolyte in electrolyte-filled twist-spun carbon nanotube muscle a new type of these artificial muscles was introduced by Lima et al. [59]. The new muscle (Fig. 1.5) is designed to be guest filled twist-spun carbon nanotube with no need of electrolyte and electrode. Paraffin as a guest material was deposited on a MWNT sheet before twist insertion and the actuation driven force is due to thermally volume expansion of paraffin. Paraffin wax is a thermally stable material with a high ability to wet carbon nanotube. Paraffin has a melting point of ~83°C with 20% volume expansion between 30° and 90°C and extra 10 % between 90° and 210°C. The 150-mm-diameter, wax-filled MWNT yarn muscle generated 10 % reversible contractile strain by Joule heating of 15 V/cm and 2.5-s period (the amount of applied current and temperature during the experiment have not been reported by authors). An average 11,500 revolutions/minute at 1200 cycles/minute and 1.38 kJ/kg work density were also produced for more than a million of cycles. Operating with high voltage and temperature as well as low contractile strain can be considered as disadvantages of these types of artificial muscles.
1.2.2. Electronic EAP (Dielectric elastomer actuators, Electrostrictive relaxor ferroelectric polymers, Liquid crystal elastomers)

The electrostatic interaction between electrodes is the simplest field-driven actuation mechanism. This mechanism is more practical in low modulus materials such as dielectric elastomers with extremely large strains of >40%. Other field-driven actuation mechanisms occur in electrostrictive relaxor and ferroelectric polymers.

1.2.2.1. Dielectric elastomer actuators

Dielectric elastomer actuators (DEAs) are well known materials that are capable of generating large strains and strain rates [60, 61]. A wide range of applications are being developed for these well studied actuators such as electroactive fluid pumps, conformal skins for Braille screens, insect-like robots and autofocus lens positioner[62, 63]. Once a voltage is applied to these materials, as a result of the attraction between opposite charges and the repulsion of similar charges, a stress generates in the dielectric which is known as the Maxwell stress[64]. This stress subsequently causes shrinkage in thickness and expansion in length direction of the dielectric material as shown in Fig. 1.6. The thickness strain $S_z$ caused by the Maxwell stress can be defined as below[63].

Figure 1.5. SEM image of a fully infiltrated homochiral coiled yarn [59].
\[ S_z = -\epsilon_r \epsilon_0 V^2 / Y z^2 \] (1.1)

Where \( \epsilon_r \) and \( \epsilon_0 \) are respectively the relative and vacuum permittivity, \( Y \) is the Young’s modulus of the elastomer, and \( z \) the thickness of the membrane. Maxwell stress is a function of applied field area and dielectric constant.

Figure 1.6. The mechanism of actuation in DEAs. The application of the voltage \( V \) between the two electrodes results in the generation of a Maxwell stress of \( \sigma \), compressing the dielectric and resulting in its lateral expansion.

Maximum contractile strains of 380% at high applied fields have been observed for these materials because of the low modulus (~1 MPa) and high dielectric strength (>100 MV/m). Generally, strains of 10-100% are achievable for these artificial muscles, which is up to two times higher than skeletal muscles. Silicone and acrylic elastomers are normally used as the most common materials in this field [65]. Safety issues are known to be a considerable disadvantage in the large devices operating with high voltages. Consequently, employing thinner sheets of elastomer or increasing dielectric constant are considered as options to
overcome safety issues by keeping the field constant and reducing the required voltage\cite{66}. These actuators are particularly suitable for devices such as robotic insects\cite{67}.

### 1.2.2.2. Electrostrictive relaxor ferroelectric polymers

Ferroelectric materials, like ferromagnets, are able to change their permanent polarisation in corresponding to dipoles that can be aligned and pass the Curie point\cite{68, 69}. Inorganic ferroelectrics, such as barium titanate are sensitive materials and change their dimensional shape in response to an electric field. However, the dimensional change (0.1\%) is significantly lower than ferroelectric polymers with strains of 10\%\cite{70}. Ferroelectric polymer actuators are fast and offer a high work density (1 MJ/m$^3$), similar to those inorganic piezo- and ferroelectrics\cite{18}. Poly (vinylidene fluoride)-based (PVDF) polymers copolymerized with trifluoroethylene, forming P(VDF-TrFE) is the most famous and capable example of these materials. The backbone of this polymer is highly polar because of the electronegativity of the fluorine, thus, field-driven alignment of polar groups generates reversible conformational changes (Fig. 1.7) which are useful for actuation movement. These materials offer elastic modulus of 0.3-1.2 GPa which, is 1000 times more than DEAs. Increasing dielectric constant and reducing thickness are alternative options to reduce the overall voltage needed (to less than 1000 V) \cite{18}.
1.2.2.3. Liquid crystal elastomers

Liquid crystal elastomers [71, 72] normally change their crystal phase and orientation in response to an applied field, temperature and light (Fig. 1.8). The most common method to generate actuation movements in these materials is to combine mesogens into either a compliant polymer backbone or use them as side chains [18, 73]. A recent study indicates that 4% strain was obtained at 133 Hz using field amplitudes of 1.5 MV/m [74]. The combination of low modulus and relatively low actuator strains, means that these materials normally offer low work density. However, a recent attempt has improved the work density by using a stiffer polymer (2% strains at 25 MV/m with a work density of 0.02 MJ/m$^3$) [75], while their performance is still far behind of relaxor ferroelectrics. Investigations to improve their performance are still at an early stage.

Figure 1.7. Reversible change of alpha and beta phases in PVDF.
1.3. Shape memory alloys (SMAs)

The shape memory effect of copper-zinc alloys (Cu-Zn) and copper-tin alloys (Cu-Sn) was firstly observed by Greninger and Mooradian in the 1950s. In general, shape-memory alloys are a group of metallic materials that offers the ability to recover a former certain length or shape when heated [76, 77]. Although, a wide range of alloys exhibits the shape memory effect, only those that can recover from a large amount of strain due to their fully reversible crystal transformation are of practical interest [78]. In the last decades, they have been utilized for military, medical, safety, and aerospace applications but their ability to recover their original shape makes them a unique candidate to use in actuators. Furthermore, their superplasticity, superelasticity and acid resistance make these materials even more technically important [77, 79]. SMAs are usually available in the form of a wire, pipes, springs or ribbons. The most common type of SMA is a nickel-titanium alloy known as NiTiNOL [80] discovered in 1959 by William Buehler and Frederick Wang at the Naval Ordnance Laboratory [76]. This SMA is known as an important candidate for smart materials and is

Figure 1.8. An example of liquid crystal elastomer (A) before stimulation (B) after stimulation.
often used in commercial applications because of its good mechanical properties, biocompatibility and shape memory effect [78]. Shape-memory alloys normally are divided into two common groups, one-way and two-way shape memory.

In one-way shape memory effect, when a shape-memory alloy is below temperature $A_s$, the metal can be bent or stretched into new shapes (Fig. 1.9). Once heated above the transition temperature the shape changes back to its original state [81]. When the metal cools again it will remain in the original shape, until deformed again. Basically, cooling from high temperatures does not lead to any macroscopic shape change and a further deformation is required to generate the low-temperature shape. The transformation phenomenon can start from $A_s$ and finish at $A_f$ (typically 2 to 20 °C or hotter, depending on the alloy or the loading conditions). $A_s$ temperature depends on the type of alloy and can vary between $-150 \degree C$ and $200 \degree C$.

In two-way shape-memory [82], the material is able to remember two different shapes: one at low temperatures, and one at the high-temperature shape. Consequently, these materials show shape changes both during heating and cooling. The material normally behaves so differently in the mentioned situations as shape memory material is able to "learn" to behave in a certain way. For example, in the normal cases, shape-memory alloys "remember" its low-temperature shape, but upon heating to recover the high-temperature shape, immediately forget the low-temperature shape. In order to keep some memory of the deformed low-temperature condition in the high-temperature phases the material can be trained. However, even a trained material is likely to lose its two-way memory effect when heated to well in excess of the transition temperature [83].
Nitinol as the most important shape memory alloys, typically consist of roughly 50 to 51% nickel by atomic percent (55 to 56% weight percent) [84]. The transition temperature is significantly depending on the composition and can change with a very small amount. The yield stress for Ni Ti can reach to 500 MPa. These materials are used in applications where the super elastic properties or the shape-memory effect are needed (actuators) because of the high cost of the metal itself and the processing requirements. These actuators are able to rapidly hold the maximum reversible strain (8%) without any permanent damage; conventional steels offer a maximum strain of 0.5% [77]. However, electrothermally driven shape-memory metal wires are expensive and suffer from hysteresis, which causes difficulty when trying to control them [85].

1.4. Twisted and coiled polymer fibres

One of the important new types of stimuli-responsive polymers are formed by twisting polymer fibres, which offer reversible change from deformed to permanent shape[92]. This unique property turns this polymer to a good candidate to be used as artificial muscles. For instance, Haines and his co-workers [85] have introduced inexpensive artificial (Fig. 1.10) muscles by using ordinary polymer fibres such as Polyethylene, Nylon 6,6 and Nylon 6. The
amount of tensile actuation was amplified by either twisting or coiling (extreme twisting) the fibers. The coiled nylon 6 muscle was able to contract 49% which is almost seven times higher than conventional shape memory alloys and two times higher than biological muscles typical contraction strain. The muscle also generated 5.3 kilowatts of mechanical work per kilogram of muscle weight, similar to jet engine with demonstrating long cycle life. The actuation control of this muscle is significantly easier in comparison to shape memory alloys. Shape memory alloys usually suffer from complex actuation control.

Figure 1.10. Carbon nanotube wrapped nylon 6, 6 monofilament (A) and (B) after coiling by twist insertion [85].

1.5. **Fluidic actuators**

Piston- cylinder fluidic actuators have shown a great capability for creating robots and tools. For instance, hydraulic cylinders mostly drive large robots and construction machines which, generate high forces however are difficult to seal, heavy and bulky, especially considering the pumps and compressors needed [17]. This type of actuators is rare in micro-devices because of fabrication issues in small scales.
Elastic or flexible fluidic actuators [93] contain at least one element that deforms elastically under the injected pressure. These type of actuators are very popular due to their easy fabrication methods with no sealing or wearing issues and are frequently being used in microactuator systems [94, 95]. McKibben artificial muscles are one of the most important and widely used type of these actuators. Table 1.3 compares performance of different types of pneumatic and hydraulic microactuators [96]. Table 1.3 also indicates that pneumatic McKibben artificial muscle exhibits the best performance to be used in linear actuators compared to other type of introduced elastic actuators. The pneumatic McKibben muscle used in this study [97] offers 6 N blocked force and 12 % contractile strain with speed of 350 mm/s. In this thesis we mainly focused in developing or creating new types of these artificial muscles.

Table 1.3. Overview comparison of pneumatic and hydraulic elastic actuators [96].

<table>
<thead>
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</tr>
</thead>
<tbody>
<tr>
<td>Operating Pressure (kPa)</td>
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<td>-</td>
<td>12</td>
<td>0.05</td>
<td>137.89</td>
<td>6</td>
<td>100</td>
</tr>
<tr>
<td>Force (N)</td>
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<td>-</td>
<td>0.01</td>
<td>0.05</td>
<td>-</td>
<td>-</td>
<td>20</td>
</tr>
<tr>
<td>Stroke (mm)</td>
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<td>0.6</td>
<td>4.5</td>
<td>0.053</td>
<td>0.084</td>
<td>40°</td>
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<td>Speed</td>
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<td>150 Hz</td>
<td>5 mm/s</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
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<td>Integrated Devices</td>
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<td>Gripper</td>
<td>-</td>
<td>Pump</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Size (mm)</td>
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<td>0.8×0.8</td>
<td>6.5×5×1.5</td>
<td>16×16×0.8</td>
<td>0.8×0.8</td>
<td>0.7×0.7</td>
<td>6×1×1</td>
</tr>
</tbody>
</table>
1.5.1. Pneumatic McKibben artificial muscles

Joseph L. McKibben was firstly introduced the McKibben artificial muscle as the most important type of pneumatic artificial muscle (PAMs) in the 1950s to assist paralysed people. Pneumatic McKibben muscle was then used as a finger driven flexor hinge splints to provide the pretension force. The Bridgestone rubber company (Japan) commercialised the idea in the 1980s under the name of Rubbertuators. The muscle normally includes an expandable elastic inner tube surrounded by a braided shell [104, 105]. The muscle usually operates with pressurized gas and the system requires a compressor as well as a gas storage container [106-109]. The pressurized air is used to increase the volume of the inner bladder and subsequently deform the braided sleeve that make up the McKibben muscle. The basic working concept of McKibben artificial muscles is that the braided sleeve translates the volumetric increase of the inner bladder to a lengthwise contraction of the braid that is capable of generating contractile forces (Fig 1.11) much greater than an equivalent hydraulic or pneumatic system. The required compressors in the conventional McKibben muscles, however, make the actuation system heavy and bulky and unsuitable to be utilized as microactuators or in portable applications where a compact size and weight minimization are desired. These type of actuators are normally easy to manufacture in a variety of sizes and also commercially available to purchase in the market.
According to Tondu et al. [104] the ideal McKibben artificial muscle can be assumed as a planar network of jointed identical pantographs as shown in Fig. 1.12. Where, $m$ columns and
n rows whose envelop is a rectangle of initial length $l_0$ and width $L_0$. The initial angle of each elementary pantograph is $\alpha_0$. It is clear that, when the network shrinks in the length direction, the initial angle moves from $\alpha_0$ to $\alpha$ and consequently the network maintains it rectangular shape. At the same time a width also increasing from $L_0$ to $L$ and a length decreasing from $l_0$ to $l$ and with assuming soft pantograph network in the form of cylinder then initial radius of $r_0$ and $L = 2\pi r$. Thus, the following equation can be proposed by assuming the side of each pantograph remains constant during the actuation.

\[
\frac{r}{r_0} = \frac{\sin \alpha}{\sin \alpha_0} \quad \text{and} \quad \frac{l}{l_0} = \frac{\cos \alpha}{\cos \alpha_0} \quad (1.2)
\]

Subsequently the contraction function is:

\[
f(\varepsilon) = \frac{1}{\sin \alpha_0} \sqrt{1 - \cos^2 \alpha_0 (1 - \varepsilon)^2} \quad (1.3)
\]

And by applying the general muscle force equation to the proposed contraction function above, we can conclude that the tensile force generated by the ideal PAM ($F_{\text{idealcyl}}$) depends upon the contraction strain ($\varepsilon = \Delta l/l_0$) as ($P =$ Internal pressure):

\[
F_{\text{idealcyl}}(\varepsilon) = (\pi r_0^2) P [a(1 - \varepsilon)^2 - b], \quad 0 \leq \varepsilon \leq \varepsilon_{\text{max}} \quad (1.4)
\]

\[
a = \frac{3}{\tan^2 \alpha_0} \quad \text{and} \quad b = \frac{1}{\sin^2 \alpha_0} \quad (1.5)
\]

As a result, the muscle normally produces the maximum force when the contraction strain ($\varepsilon$) is zero as below:

\[
F_{\text{idealcyl max}} = (\pi r_0^2) P(a - b) \quad (1.6)
\]
Based on above simple equation, generated force is a function of initial angle of the braided sleeve, internal pressure and muscle radius. Figure. 1.13 shows the dependency of the generated force on initial angle and internal pressure. It appears that, the generated force decreases significantly with increasing the initial angle up to critical angle which is $54.44^\circ$ and then the muscle produces negative forces which can be interpreted as an expansion instead of contraction. It was found that the muscle generates higher forces for the same initial angle and radius with increasing the amount of internal pressure. The difference between red and black lines is more significant in the lower initial angles.

![Graph showing the relationship between maximum generated force (at zero strain) and initial angle of the braid for PAMs with a starting radius of 1 mm and pressurized to either 0.40 or 0.55 bar.](image)

Figure 1.13. The relationship between maximum generated force (at zero strain) and initial angle of the braid for PAMs with a starting radius of 1 mm and pressurized to either 0.40 or 0.55 bar.

It is also important to note that according to the equation 1.4 the muscle generates the highest contraction strain when the generated force is 0:

$$
\varepsilon_{\text{ideal cyl max}} = 1 - \left( \frac{1}{1.732 \cos \theta_0} \right)
$$

(1.7)
Equation 1.7 indicates that the amount of contraction strain of the muscle only depends on initial angle of the braided sleeve and is independent of internal pressure. Figure 1.14 shows the dependency of contraction strain on initial angle of the braided sleeve. The amount of contraction strain reduces with increasing the initial angle (similar to the force trend in equation 1.6) and reaching zero contraction strain at critical angle (54.44°). The behavior of the muscle changes dramatically above the critical angle and produces expansion strains, the phenomena that also were observed in force behavior (equation 1.6). This behavior proves that McKibben artificial muscles can be adjusted for specific applications where either expansion or contraction strains are required.

Figure 1.14. The relationship between contraction strain and initial angle of the braid.

This particular artificial muscle, however, presents some disadvantages such as the requirement of a separated mechanical air compressor, a noisy system, a heavy system to carry for human or robots and high electricity consumption [109]. To overcome mentioned disadvantages several attempts, have recently been made to replace the air with water or chemo-sensitive materials to introduce more compact and less noisy system.
1.5.2. Hydraulic McKibben artificial muscles (HAMs)

Hydraulic McKibben muscles have been introduced recently as a new generation of McKibben muscles, which operate with pressurized water or oil instead of pressurized air [96, 111]. According to Tiwari et al. [107] and Meller et al. [106], the use of bulky compressors can be avoided in hydraulic artificial muscles (HAMs), thus making compact design possible by using small pumps. Meller et al. [106] have also clearly demonstrated that the HAMs have approximately doubled the energy conversion efficiency of PAMs. Moreover, it has been shown that it was possible to design relatively ‘large’ hydraulic McKibben muscles to actuate human limb-size robots, or even very powerful ones thanks to the use of a particular strong external braided sleeve [108]. It was also practical to develop microscale HAMs like the ones proposed by Moon et al. [109], or by Solano and Rotinat-Libersa [112] for millimeter scale robot development. Table 1.4 summarizes the performance of previously introduced HAM systems. The reported systems vary considerably in size and operating pressures. The reported maximum (blocked) forces covered a wide range with the larger diameter muscles generated the higher forces. Table 1.4 indicates that the response time of the hydraulic systems is longer than pneumatic systems, as a result of the higher viscosity of the water. The higher viscosity of water ultimately consumes more time to fill up the inner bladder and stimulate the braided sleeve.
Table 1.4. Performance comparison of HAMs reported in the literature. Maximum values are shown for stroke and blocked force and minimum values for response time.

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<tbody>
<tr>
<td>Muscle length (mm)</td>
<td>160</td>
<td>700</td>
<td>237</td>
<td>61</td>
<td>173</td>
</tr>
<tr>
<td>Diameter (mm)</td>
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<td>40</td>
<td>2</td>
<td>1.5</td>
<td>32.3</td>
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<tr>
<td>Braid angle (°)</td>
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<td>-</td>
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<td>28.7</td>
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<td>Muscle weight (kg)</td>
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<td>0.02</td>
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<td>9V electric pump</td>
<td>Motor and piston</td>
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<td>Free stroke (%)</td>
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<td>Blocked force (N)</td>
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<td>6</td>
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<td>Response time (sec)</td>
<td>-</td>
<td>-</td>
<td>2.8</td>
<td>-</td>
<td>-</td>
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</table>

1.5.3. pH-activated McKibben artificial muscle

In an alternative approach, Tondu and his co-workers [113-116] have also developed a new generation of McKibben muscles by replacing pressurized air with pH sensitive hydrogel spheres. These reactive chemical products were placed inside the McKibben muscle inner tube and flow systems were used to generate its swelling and de-swelling by passing basic and acid solutions, respectively, through the inner tube (Fig. 1.15). pH sensitive hydrogels seemed very promising in generating reversible swelling and de-swelling behavior (Fig. 1.16). It is, however, important to note the fact that this functioning principle assumes that the swelling phenomenon does not inhibit the circulation ability of acid and alkaline solutions through the inner tube of the artificial muscle. As a result, a thin natural rubber inner tube is surrounded by a nylon braided sheath attached to both ends in such a way that solutions can circulate from one end to the other. The initial active length is around 100 mm; the initial
external diameter is around 8 mm and initial braid angle is around 25°. These dimensions were chosen in accordance with swelling properties of tested materials.

Figure 1.15. Swelling principle of the Tondu's pH-muscle: (A) macroscopic view of the artificial muscle inner chamber during the diffusion process of a NaOH solution: (B) mechanism of Na ions fixation by an ion-exchange resin balls in its acidic form (COOH) [116].

Figure 1.16. Scheme of the experimental set-up of Tondu's muscle in isometric and isotonic conditions [116].
Isotonic contraction of Tondu’s pH-muscle was performed against loads between 0.5 and 3 kg as shown in Figure 1.16. After 90 minutes time response, the muscle shows maximum contraction ratios between 15% and 18.5% for all range of loads (0.5, 1, 2, 3 kg). However, the contraction percentage reduces by load increasing; none of the samples were able to achieve 20% contraction that is achievable with biological muscles. Furthermore, the isotonic response time (90 min) is dramatically high compared to biological muscle, which is well below one second. As a result, further investigation to increase the contraction ratio and reduce the response time of this muscle to mimic real muscles is necessary. The muscle also generated 118 N isometric forces after 9 minutes, which is 10 times faster than of its isotonic response time (90 min). Literature reviews [97, 117, 118] indicate that, important parameters such as type of sensitive material, initial braid angle of the braided sleeve, size and geometry of the sample and mechanical properties of the surrounded inner bladder [3] and braid can directly affect the amount of contractile stain, force generated and time response of this particular muscle. This muscle also suffers from high device packaging as a result of the piping and pumps which makes the system unsuitable to be utilized in microactuators systems.
1.6. Actuators comparison and thesis aim

Table 1.5. Advantages and disadvantages of prominent artificial muscles in comparison to biological muscles.

<table>
<thead>
<tr>
<th>Artificial muscle</th>
<th>Advantages</th>
<th>Disadvantages</th>
</tr>
</thead>
</table>
| Conducting polymers | Low operation voltage (~2V)  
High strain (40%) | Slow actuation rate  
(<1%/s) |
| CNT actuators | Very fast (<10ms)  
High work density (1MJ/m$^3$)  
Temperature stability (>450 °C) | Low contractile strain  
(<2%) |
| Torsional Carbon Nanotube Artificial Muscles [13] | Excellent torsional actuation  
(15000°)  
Very small size  
(15µm in diameter) | High device packaging  
Low contractile strain  
(< 3.4%) |
| Electrically, chemically, and photonically powered torsional and tensile actuation of hybrid carbon nanotube yarn muscles [59] | Compact actuation system  
Good torsional actuation  
(11500°) | Low contractile strain  
(< 11%)  
Need for guest material  
Moderate voltage operation (15V/cm) |
| Dielectric elastomers | Large strain (10-300%) and strain rate | High voltage operation  
(>1 Kv) |
| Relaxor ferroelectric | Very fast strain rate (>2000%/S)  
high work density (>1000kJ/m³) | Low strain (1-5%)  
High voltage (>1000V) |
| Liquid crystal elastomers | Sensitive to applied field | Low modulus (>100 MPa)  
Low strain (±4%) |
| Shape memory alloys | Biocompatibility  
Good mechanical properties | Complex actuation control  
Expensive  
Low strain (7%) |
| Artificial muscles from fishing Line and sewing thread | Excellent contractile strain (49%)  
Very compact system  
Excellent mechanical work (5.3) | Limited operating temperature  
Creep problem |
Table 1.5 represents the advantages and disadvantages of the most common types of available artificial muscles in order to compare them with biological muscles. It clearly appears that, CNT actuators, shape memory alloys and relaxor ferroelectric are very fast actuators but offer very low strains around 1-5%. Furthermore, conducting polymers are not fast; however, low voltage operation still makes them an interesting candidate to be employed as an artificial muscle when the low electricity efficiency is tolerable. Foroughi et al. [13] and Lima et al. [59] muscles offer very practical torsional actuations but suffer from low contractile strains, which makes them inappropriate where linear biological like muscle behavior is required. Haines et al. [85] muscle is a very practical device with properties very close to that of linear biological muscle. Further investigation is currently taking place in order to overcome the disadvantages of this muscle such as creep.

Pneumatic/hydraulic artificial muscles are fast (1 sec – 1 min), show high range of strains (25-30%) and isometric forces, although, these systems are heavy and bulky which makes them
an unsuitable option when a compact and light system is required such as microactuator systems.

Tondu et al. muscle [113-116] generates reasonable actuation strain and force. However, there are still some remaining problems that need to be considered, such as the long response time (> 10 min), and the required pump for delivering acid/base solutions to the pH sensitive hydrogel.

The main aim of this thesis is to create or develop a new type of McKibben artificial muscle by eliminating the need of pump/compressor as well as piping to reduce the device packaging of these muscles. The lighter and more compact type of these muscles is more suitable for portable applications. Attempts to improve the performance of the currently available hydraulic McKibben muscles by reducing the stiffness of the inner tube have also been included in this study. It has been found that this part of the muscle consumes some of the input pressure and ultimately reduces the muscle performance. Paraffin as a temperature sensitive material was also selected to replace the pressurized fluid used in HAMs. A thermodynamic equation was subsequently introduced to predict the performance of these muscles by using temperature as an actuation driving force.
1.7. References


Chapter TWO

The effect of geometry and material properties on the performance of a small hydraulic McKibben muscle system

This chapter presents the study that has appeared in the publication:

2.1. Introduction

As discussed in chapter 1, McKibben artificial muscles are one of the most popular biomimetic actuators, showing similar static and dynamic performance to biological muscles[1-4]. In particular, their pneumatic version offers high force to weigh ratio, high speed and high strain in comparison to other actuators [5, 6]. This particular artificial muscle, however, presents some disadvantages such as the requirement of a separated mechanical air compressor, a noisy system due to exhaust during depressurization, a heavy system to carry for human or robots and high electricity consumption[7]. One of the options to overcome those disadvantages is to replace the pressurized air with pressurized water or oil in order to manufacture a fully sealed system with no requirement of a refiling process[8]. Hydraulic McKibben muscles have been introduced recently as a new generation of these muscles [9-13].

In this chapter we introduce a cost-effective (US$ 2), fast and small hydraulic McKibben muscle (35–80 mm resting length) that offers 23% actuation stroke (under 4.9 N isotonic load) and up to 26 N isometric force generated in about one second. The muscle simply operates with pressurized water that is supplied from a small container (25 ml) in a fully sealed system via small pump that can be easily carried in portable applications. The electricity consumption of the pump is significantly low (6 V, 0.14 A) and operates with small batteries. This unique actuation system is lightweight and can be easily modified to be employed in small robotic systems where large movements in short time are required. The effect of muscle length and the stiffness of the inner tube (bladder) on muscle performances such as force generation, contraction ratio and response time have also been experimentally investigated. The effect of bladder stiffness has also been modified as described in the next section.
2.1.1. Effect of the inner tube stiffness on the static properties of an ideal cylindrical McKibben artificial muscle

A classical model of an ideal purely cylindrical McKibben artificial muscle relates the static force $F$ produced by the muscle to its control pressure $P$ and its contraction strain $\varepsilon$\cite{14}. This model can take the following form:

$$F_{\text{ideal cyl}}(P, \varepsilon) = (\pi R_0^2)P \left[a(1 - \varepsilon)^2 - b\right], \quad 0 \leq \varepsilon \leq \varepsilon_{\text{max}}$$ (2.1)

where $R_0$ is the initial radius of the braid, $a = 3/\tan^2 \alpha_0$, $b = 1/\sin^2 \alpha_0$. Equation 2.2 indicates that the maximum contraction strain is independent of applied pressure and is given by:

$$\varepsilon_{\text{ideal cyl max}} = 1 - \left(\frac{1}{1.732 \cos \alpha_0}\right)$$ (2.2)

This equation only requires the knowledge of two geometric parameters characterizing the artificial muscle: the initial braid angle $\alpha_0$ and the initial muscle radius $R_0$ which is usually assumed to be the initial external inner tube radius, and considered as being equal to the initial internal braided sleeve radius. This model assumes a full transmission of the pressurized stress inside the inner rubber tube to the external braided sleeve. Such an assumption is generally verified in the case of pneumatic artificial muscles working in a typical [1–5 bars] range if a sufficiently thin inner tube made of a soft rubber was chosen. Meller et al. [15] and Pillsbury et al. [16] have recently demonstrated the effects of bladder stiffness on HAM/PAM performance with increasingly stiff bladders significantly limit the maximum strain achieved and slightly reducing the blocked force. A semiempirical approach
was introduced by Meller et al. [15] to account for bladder stiffness in the ideal model by introducing fitting parameters $K_F$ and $K_\varepsilon$ for force and strain, respectively:

\[ F = K_F(\pi r_0^2) P \left[ a(1 - K_\varepsilon \varepsilon)^2 - b \right] \]  

(2.3)

where $K_F$ is the ratio of the measured blocked force to the maximum force predicted by Eq. (2.1) and $K_\varepsilon$ is the ratio of the maximum strain predicted by Eq. (2.1) to the measured maximum strain. In this chapter the effect of bladder stiffness on HAM performance is reconsidered and a simple alternative approach to modelling.

2.2. Experimental

2.2.1. Prototype fabrication

McKibben artificial muscles normally consist of four important parts: an elastic tube, a cylindrical reinforcement braid, and two connectors for the pressurized fluid supply. To determine the effect of inner tube stiffness on the actuator performance, inner tubes made of natural rubber latex with two different thickness (0.28 and 0.56 mm) and a silicon rubber tube (thickness of 1 mm, Holman Flex tube) were used inside the braid. All bladders had an external diameter of 4.5-5.0 mm. Shimadzu EZ tensile machine were also used to measure the stiffness of the bladders by axially stretching the rectangular bladder samples. Cylindrical braids with thickness of 0.44 mm and outer radius in the unstrained state ($R_0$) of 3 mm, made from polyphenylene sulfide (PPS) were obtained from JDD TECH Company, China. Crescent zip lock was also used to connect the muscle to tube connections. The initial angle ($\alpha_o$) of the braided sleeve (Fig. 2.1) was kept constant at 35°. The muscle was fabricated as
follows: first, the inner tube was cut into the desired length of 35, 50, or 80 mm. Next, the inner tube was inserted into the braided sleeve, and finally the PVC connector tubes were glued to both ends of the inner tube.

Figure 2.1. Polymeric braided sleeve used for hydraulic McKibben muscle. (a) resting state (b) expanded state (c) photographs of HAMs of different lengths with tube connectors.

**2.2.2. Actuation set up**

The experimental set up was specially designed to measure actuation strain, generated force, response time and water pressure (Fig. 2.2). The sealed actuation system consisted of four important parts: a low voltage water pump (6 V, flow rate: 0.5 l/min, Flodos/NF6 KPDCB), a small water container (25 ml), McKibben muscle and a manual valve. Care was taken when filling the actuator and connecting tubes with water so as to remove any trapped air. The
entire system had a total weight of only 350 g, making it easily usable in robotic machines. The main mass of the system was due to the pump and power supply which can be optimized depending on the application. The operation is simply by opening and closing the manual valve. When the valve is closed while the pump is working the water becomes pressurized inside the muscle and causing the muscle to contract in length. Dual-mode lever system machine (Aurora Scientific, Model 300B) and Shimadzu EZ tensile machine were employed to record actuation stroke and force generated. The lever arm and tensile machine were connected to the artificial muscle, while the other end of the muscle was fixed. An e-corder data logger (ED 410, e-DAQ) was used to connect the lever arm unit to a PC, and e-DAQ Chart was used to record the data. The internal water pressure inside the muscle was also monitored by using a digital pressure meter (GEMS sensors and controls-3300R012). The current and voltage applied to the pump were noted periodically.
In this chapter, both isometric and isotonic standard tests for hydraulic artificial muscles were performed to obtain actuation results in accordance with corresponding physiological definitions:

*Isometric force:* The muscle length was fixed to be constantly equal to its initial length and the maximum muscle force generation (the ‘blocked force’) recorded by using a force sensor as shown in Figure 2.3.

![Figure 2.3. Isometric test (a) before muscle stimulation (b) after muscle stimulation.](image_url)
Isotonic contraction: A given tensile force of 4.9 N was applied to the muscle by using a lever arm. This method was used to record the isotonic length variation of the muscle as described in schematically in Figure 2.4.

(a) \[ F_1 = 0 \]
(b) \[ F_2 = F_1 \]

Figure 2.4. Isotonic test (a) before muscle stimulation (b) after muscle stimulation.

Force–stroke curves: The possible force/stroke combinations were obtained by first measuring the isometric blocked force and then allowing the muscle to contract while simultaneously measuring force and stroke at a fixed pressure.

2.3. Results and discussion

2.3.1. Effect of the unloaded muscle length on isometric force generation and isotonic actuation strain with constant pressure

An analysis of the effects of actuation length on the response time and isotonic strain behavior was made by comparing three different muscles lengths of 35, 50 and 80 mm under load of 4.9 N and water pressure of 2.5 bar (250 kPa) applied for ~1 s and then released. Figure 2.5 exhibits that all muscles contracted continuously during the pressurization period.
with the shortest length (35 mm) achieving a strain of 23% in about 1 s, while the muscle with lengths of 50 and 80 mm generated smaller actuation strains of 18.5 and 16%, respectively. The time to reach a target strain of 15% increased with starting muscle length from 0.5 s (35 mm) to 0.7 s (50 mm) and 0.9 s (80 mm). The faster response seen in the shorter muscle was likely attributed to the smaller volume of water needed to pressurize the muscle. The expansion response time due to depressurizing the muscles was also dependent on muscle length but considerably faster than pressurization in all cases. Furthermore, the muscle of 80 mm produced the highest actuation displacement and greatest power (0.075 Watts) compared to the other two muscles with shorter lengths.

Figure 2.5. Isotonic actuation test under constant water pressure (2.5 bar) applied for 1.1 sec and given load of 4.9 N; pressurization-depressurization tests were performed four times on each HAM and the average maximum strains for each HAM length are shown by solid squares. The ranges of maximum strain values are represented by the error bars calculated as one standard deviation around the mean.

The instantaneous power was also calculated during contraction as the product of displacement and load per contraction time and is shown in Figure. 2.6. It appears that the
longest muscle (bladder stiffness: 78 N/m) produced the highest power around 0.075 W which peaked after just 0.32 seconds, as a result of generating more displacement. All of the muscles were able to produce 0.052 W power after 0.19 seconds regardless of their lengths. The overall power conversion efficiency was 8.9% based on the input electrical power of 0.84 W needed for the hydraulic pump. The efficiency of this particular system is higher than liquid crystal elastomers (<5%), conducting polymers (1-5%), carbon nanotube actuators (0.1%) and shape memory alloys (5%). Biological muscles (8-40%) and dielectric elastomers (30%) offer significantly more efficiencies than this muscle.

Figure 2.6. Corresponding power output obtained from the isotonic test; (2.5 bar) and given load of 4.9 N.

Isometric force generation is one of the essential requirements for many applications of artificial muscles such as robotic surgery and artificial jumping legs. According to Volder et al. [17] high forces of 1–10 N are required for robotic devices and surgery tools. Here we compared the three different HAM muscles with water pressure of 2.5 bar to investigate
the effect of actuation length on muscle performance. The maximum isometric force of 26 N was recorded for the 80 mm long muscle in just 1.4 s. According to Figure. 2.7, the muscle ability in force generation scales approximately with actuator length. The shortest muscle (35 mm) generates isometric force of 11 N in 1.2 s, which is almost half of the muscle with 80 mm length. These results were very consistent for four consecutive pressurization–depressurization tests. Only one muscle was made at each length and each muscle was tested using four separate pressurization /depressurization steps to assess the reproducibility in force generation.

Previous studies have shown little effect of braid length on the force generated when pressurized [18]. However, our muscles have comparatively small aspect ratios and below the recommended ratio of 14 [19] so that end effects may limit the force generated, especially in the shorter samples. The decision on making samples with small aspect ratios was made because of the interest in microactuator applications. The performances of our muscles are compared with previous HAMs systems, as summarized in Table 1.4 (chapter 1). The reported systems vary considerably in size and operating pressures. The reported maximum (blocked) forces covered a wide range with the larger diameter muscles generated the higher forces. Three previous studies used similarly small diameter braids as used in the present work of less than 6 mm [7, 12, 13]. The maximum contraction (free) strains from these small diameter braids were of a similar magnitude (~20%) and the maximum blocked forces were either similar or lower than those reported in the present study. The comparison highlights that it is possible to generate HAM performance comparable with other literature studies with the use of a low voltage/low power electric pump and a limited pressure range of 2.5 bar.
Figure 2.7. (A) Isometric force test under constant water pressure (2.5 bar). Pressurization-depressurization tests were performed four times on each HAM and the average maximum forces for each HAM length are shown by the solid squares. The ranges of maximum force values are represented by the error bars calculated as one standard deviation around the mean. (B) Isometric force test under constant water pressure (2.5 bar). The dot points are indicating the maximum number of each pressurization-depressurization cycle which was obtained four times on each HAM.
2.3.2. Effect of the inner tube stiffness on hydraulic McKibben artificial (HAM) muscle performances

Further experiments focused on the effect of the bladder on the pressure needed to develop an isometric stress in the range of 1–4 N and the corresponding free stroke. Three different individual muscles with the same geometry (35 mm long and 6 mm diameter braid) were made with three different inner tube stiffnesses of 78, 150 and 490 N/m. The data of (Fig. 2.8 c) shows that the muscle with the stiffest inner tube needs dramatically more water pressure (2.14 bar) to reach the targeted static force of 1.2 N and exhibits just 2.5% contraction free strain. In contrast, the muscle with smallest inner tube stiffness (78 N/m) needs only 0.33 bar pressure for the same amount of force and shows 4% contraction strains (Fig. 2.8 a). In these experiments a targeted force was fixed and once the muscle reached to the targeted force the amount of input pressure was recorded in order to compare with other muscle made with different inner tube stiffness. All tested samples showed a similar degree of hysteresis in the loading and unloading curves that has been attributed previously to braid friction [14]. The hysteresis directions are similar for all of the other pressures in the same graph.
Figure 2.8. Typical static forces and contraction strains produced by a hydraulic artificial muscle emphasizing the role of input pressure and illustrating the hysteresis phenomenon and the fundamental role of the stiffness of the bladder. The muscles with initial length of 35 mm and inner tube stiffness of (a) 78 N/m (b) 150 N/m and (c) 490 N/m. Note the noise seen in (c) is from pump vibration that is more prominent at the high pressures used for this sample.

The above results confirm that HAM performance is influenced by parameters other than the initial braid angle and radius, as suggested by Eq. (2.1). All contraction strains were less than 12% which is well below the prediction of 30% from Eq. (2.2) based on a starting braid fiber angle of 35°. Blocked forces for a given input pressure were also considerably lower than the predicted values. Increasing bladder stiffness tended to decrease the blocked force and contraction strain, as a result of the elastic deformation of the bladder and friction between the bladder and braid.

A second set of experiments was performed to further investigate the effect of bladder stiffness on HAM performance. Longer samples lengths of 80 mm were used to reduce end-effects that diminish achievable contraction strains. Three different bladders were used and the results shown in Fig. 2.9 demonstrate that contraction ratios now approach the predicted value of 30%, at least at the higher input pressures. For example, the HAM using the least
stiff bladder and pressurized to 2.5 bar gave a maximum contraction ratio of 28%. However, even with these longer samples, it is clear that the increasing bladder stiffness reduces both the blocked force and contraction ratio for a given input pressure.
To quantify the effect of the bladder stiffness on HAM performance, we consider that the elastic resistance due to the inflation of the bladder material tends to reduce the pressure available to work against the braid and restricts the McKibben muscle output. Thus, we offer a simple approach to account for the pressure needed to inflate the bladder. We use the ideal McKibben muscle relation of Eq. (2.1) but modify the input pressure by subtracting two contributions: the threshold pressure needed to inflate the bladder to make contact with the braid ($P_{th}$); and the pressure needed to elastically deform the bladder when in contact with the braid and causing HAM contraction. The threshold pressure is a fixed term that depends on the bladder and braids geometries and the bladder stiffness and occurs when the un-inflated
bladder outer diameter is smaller than the braid inner diameter. Figure 2.10 shows the measured blocked force at several input pressures for the three different bladders used in the present study. Also shown is the theoretical blocked force obtained from Eq. (2.1). These results show that the variation in blocked force with pressure are similar for all bladders and correspond quite closely to the expected trend. However, the measured results are offset along the pressure axis by an amount that increases with bladder stiffness. Extrapolating the experimentally measured values for each bladder material to zero force provides an experimental measure of the threshold pressure, $P_{th}$. These values are given in Table 2.1.

![Figure 2.10](image-url)

Figure 2.10. Static (blocked) forces measured at various input pressures for HAMs made with 3 different bladder materials: circular and triangular symbols are measured values. The linear line with square symbols is the theoretical values obtained from Eq. (2.1) that assumes and ideally thin bladder ($R_0 = 3\, \text{mm}, \alpha_0 = 35^\circ$).
Table 2.1. Threshold pressure and parameters for each bladder.

<table>
<thead>
<tr>
<th>Physical information of bladder materials</th>
<th>Stiffness 78 N/m</th>
<th>Stiffness 150 N/m</th>
<th>Stiffness 490 N/m</th>
</tr>
</thead>
<tbody>
<tr>
<td>Threshold pressure $P_{th}$ (bar)</td>
<td>0.08</td>
<td>0.21</td>
<td>1.89</td>
</tr>
<tr>
<td>Elastic modulus (MPa)</td>
<td>0.9</td>
<td>0.9</td>
<td>0.8</td>
</tr>
<tr>
<td>Inner diameter (mm)</td>
<td>5.1</td>
<td>4.5</td>
<td>3.0</td>
</tr>
<tr>
<td>Outer diameter (mm)</td>
<td>5.6</td>
<td>5.6</td>
<td>5.0</td>
</tr>
<tr>
<td>Wall thickness (mm)</td>
<td>0.28</td>
<td>0.56</td>
<td>1.0</td>
</tr>
</tbody>
</table>

The stiffnesses of the bladders were also measured by using tensile testing machine. Rectangular sheets were obtained from each bladder material by cutting the bladder tubes open. The sheets were ~ 9 mm wide and 20 mm long. The sheets were axially stretched up to the breaking point in order to plot force vs stroke curves. The slope of the linear part of force versus stroke curves exhibits the stiffness for each individual bladder as shown in Fig. 2.11.
Figure 2.11. Typical force vs stroke diagram obtained by tensile testing machine of three different bladder materials.

For pressures below the threshold the bladder is not yet in contact with the braid and no force is generated by the muscle. The measured input pressure was also adjusted by subtracting the pressure needed to elastically deform the bladder ($P_{el}$) when in contact with the braid during HAM contraction. This pressure was estimated for simplicity by the standard mechanics analysis of a pressurized cylinder where the circumferential strain in the bladder is given by:

$$\varepsilon_b = \frac{\Delta r}{r_0} = \frac{Pr_0}{E_b t_b}$$

(2.4)

Where $P$ is the internal pressure; $r = r - r_0$ with $r$ and $r_0$ representing the pressurized and unpressurized inner radius of the bladder; and $E_b$ and $t_b$ are the bladder elastic modulus and wall thickness, respectively. When $P$ reaches the threshold pressure ($P = P_{th}$), the bladder
makes first contact with the braid. We assume that for pressures in excess of the threshold ($P > P_{th}$), the bladder is in contact with the braid so that outer radius of the bladder ($r + t_b$) is the same as the inner radius of the braid ($R_0$). The change in braid radius is then estimated by:

$$\Delta R = \frac{r_0^2}{E_b t_b} (P_{el} - P_{th})$$

(2.5)

Where $P_{el}$ is the pressure needed to further inflate the bladder so that it maintains contact with the braid during HAM contraction. The change in braid length can be determined using the geometry relations appropriate for the helically wound fibers in the braid:

$$R = \frac{bsin\alpha_0}{2\pi N} \quad \text{and} \quad L = bcos\alpha_0$$

(2.6)

Where the braid of length $L$ is constructed from inextensible fibers of length $b$ and wrapped $N$ times at an angle of $\alpha_0$ to the braid long axis. Combining these equations gives the following relation that shows that the pressure needed to elastically expand the bladder increases with increasing HAM contraction strain.

$$P_{el} = P_{th} + \frac{E_b t_b b}{2\pi N r_0^2} \left[ \left(1 - \frac{L^2}{L_0^2}cos^2\alpha_0\right)^{\frac{1}{2}} - sin\alpha_0 \right]$$

(2.7)

Where $L_0$ and $\alpha_0$ are the starting braid length and fiber wrap angle. The approach is to use the ideal McKibben muscle relation of Eq. (2.1) modified as:
\[ F_{\text{idealcylic}}(P, \varepsilon) = (\pi R_0^2) P^* [a(1 - \varepsilon)^2 - b], \quad 0 \leq \varepsilon \leq \varepsilon_{\text{max}} \] (2.8)

With the input pressure \( P^* = P_{\text{applied}} - P_{\text{el}} \) where \( P_{\text{applied}} \) is the actual applied pressure that is modified by subtracting the pressure needed to elastically deform the bladder \( (P_{\text{el}}) \) calculated from Eq. (2.7).

Fig. 2.12 indicates that the calculated values from this modified model (Eq.13) are in reasonable agreement with the measured values. In contrast, the standard McKibben muscle relation of Eq. (4) does not include the effect of bladder stiffness and, as shown by the dotted lines in Fig. 2.12, this standard model over-estimates both the force and strain generated when stiffer bladders were used. The subtraction of the threshold pressure in the modified model ensures that the calculated blocked maximum force is in closer agreement with the measured values, since the unmodified pressure greatly over-estimates the achievable force when stiffer bladders were used. The subtraction of the pressure needed to elastically deform the bladder during HAM contraction can also significantly reduce the maximum contraction strain. The theoretical prediction using unmodified pressure predicts the same maximum strain regardless of the bladder material used or input pressure. The measured values clearly show that the maximum strain increases with increasing input pressure and is affected by the bladder stiffness. These effects are modelled using the simple approach applied here. While the modified model captures the main trends, there remain discrepancies between the calculated and measured force / strain values. The modified model is based on material linear elasticity and assumes a constant bladder wall thickness. Both assumptions are not strictly valid for the large strains occurring in the elastomeric bladders used with the HAMs. However, incorporating non-linear elasticity effects adds significant complexity to the model and was not considered in the present study.
Experimental data for bladder stiffness
- 150 N/m
- 78 N/m

Modified model

Standard McKibben equation
Figure 2.12. Comparison of measured and calculated force-strain curves for 80 mm long HAM pressurized to (a) 0.66 bar; (b) 1.5 bar and (c) 2.5 bar. Calculated values using the standard McKibben muscle relationship (Eq. (2.1)) are shown by the dotted lines. Experimentally measured values and values calculated using the modified model (Eqs. (2.7) and (2.8)) are shown by the symbols and dashed lines, respectively, where the squares are for the 78 N/m bladder; triangles represent the 150 N/m bladder; and circles are for the 490 N/m bladder (part c only).

As mentioned earlier, Meller et al [15] has also introduced a semiempirical model, which is able to accurately predict the amount of generate force as well as contraction strain by taking to account two important fitting parameters $K_F$ and $K_\varepsilon$. However, these parameters were obtained from experimental data and therefore are only suitable for Meller’s experimental conditions. Meller et al [15] has also constructed a new type of McKibben muscle by using LDPE bladder which offers very similar performance in comparison to predict data of Equ.1.2. It is because the LDPE bladder is completely attached to the braided sleeve inside the muscle; therefore no pressure was needed to consume to make this attachment (Consuming pressure to make this attachment is a normal process for conventional McKibben muscles with rubber bladder). This consumed pressure can ultimately affect the accuracy of
the equation 2.1. LDPE bladder as a plastic material deforms and consequently consumes no pressure in this regard, and behaves similarly to assumption of equation 2.1. These two factors confirm that our approach in introducing new equation is correct.

2.4. Conclusions

Hydraulic McKibben artificial muscles are easy to manufacture, and perform quite similarly to biological muscles in terms of response time, isotonic actuation strain and isometric force generation. In the present study, HAMs were scaled down to a diameter of 6 mm and lengths of 35–80 mm, which makes them more suitable for in applications such as robotic fingers for surgical tools. Forces up to 26 N were achieved at a pressure of 2.5 bar, with an overall system response time of just 1.4 s. Actuation strain of 23% was obtained in just 1.1 s with given load of 0.5 kg and 2.5 bar. It has been found that the effect of stiffness of the inner tube on muscle performances is considerable and should be carefully chosen. In particular, the pressure needed to inflate the bladder to make contact with the braid reduces the maximum force achievable. Secondly, the pressure needed to elastically deform the braid during HAM contraction reduces the maximum achievable strain. A simple method was proposed to quantitatively estimate these effects with reasonable accuracy. The best performing HAM muscles at low pressures generated by low-voltage portable water pumps were achieved with the least stiff bladders.
2.5. References

Chapter Three

3D printed braided sleeve to be utilized in fabricating McKibben artificial muscles

This chapter presents the study that is in preparation as a manuscript:

Sangian D, Jeirani A, Naficy S, Beirne S, Spinks GM. 3D printing braided sleeve using biocompatible PCL polymer to be utilized in fabricating McKibben artificial muscles.
3.1. Introduction

As mentioned in chapter 1, McKibben artificial muscles are one of the most practical artificial muscles because of their large blocked forces, high contraction strains and fast response time. They also exhibit very similar performances to biological muscles and are widely used in robotic tools. These muscles normally consist of two main parts: an inner bladder and a braided sleeve. The volume change of the inner bladder acts on the braided sleeve causing the entire muscle either to shrink or expand in the length direction depending on initial angle of the braided sleeve. The previous chapter investigated the influence of the inner bladder stiffness on the performance and modelling of McKibben artificial muscles and in this chapter, the influence of the braid is considered.

The braided sleeve used in conventional McKibben artificial muscles are sourced commercially and manufactured with industrial braiding machines. The braiding machine assembles the individual fibers by using several rotary spools to produce a cylindrical hollow braided sleeve. However, conventional braiding machines suffering from three important disadvantages. Firstly, producing consistent cover factor in single product for high friction fibers is limited due to the friction between fibers. Secondly, the braiding machines have some restrictions such as limitation in generating a wide range of braid initial angles as well as constraint in small, research scale production especially when only short lengths of experimental fibers are available. Thirdly, producing braided sleeve with connected fibers in junction points is not possible with available braiding machines. As a result, an alternative method that can overcome the mentioned disadvantages is desired.

In this chapter we attempted to investigate the possibility of introducing a more versatile method to produce the braided sleeve, particularly for producing short length samples as well as achieving connected fibers in junction points. The braided sleeves in this study were made
using KIMM Bioplotter 3D printing machine. Each individual line made of polycaprolactone material was carefully printed around a cylindrical steel rod. The braided sleeves were made in two different ways: connected and disconnected fibers in the junction points in order to investigate the effect of this phenomenon.

### 3.2. Experimental

#### 3.2.1. Fabrication of Braided Sleeves

To investigate the effect of the connection or disconnection between crossover points (Fig. 3.1), on muscle performance two different types of braided sleeves with similar geometry were produced. The KIMM Bioplotter machine was used to print melt-extrudable polymer (polycaprolactam: PCL) around a polished steel rotating mandrel. Two rings were also printed and connected to the ends of the braided sleeve to prevent the fibers from unravelling (Fig. 3.2). Polycaprolactone with a molecular weight of 45000 and melting temperature of 60°C was also used as the braid fiber material. A nozzle diameter of 400 µm was used with an extrusion pressure of 100kPa.

(a)
Figure 3.1. (a) the schematic view of braided sleeve indicating the junction point (b) deformed shape of one diamond after pressurization, disconnected junction point (left) connected junction point (right).

Figure 3.2. Photograph of printing set for producing polymeric braided sleeve.
The braided sleeve with connected crossover points (M₁) was simply manufactured by continuous printing onto a clockwise rotating mandrel. The printing was firstly performed right to left to make one helix of the braid and then print direction was reversed to form the second, overlapping helix. In this process each individual fibre was printed on top of each other and strong connections formed at the crossover points because of localised melting and solidification. Finally, the braided sleeve was carefully removed from the steel mandrel.

To produce a braid with disconnected crossover points (M₂), the printing process was divided into two different sections. The right to left printing direction was firstly performed as explained above and then the sample was dip coated in alginate solution. Once the alginate solution was completely dried, the left to right printing direction was performed. The sample was then immersed in the water to dissolve the alginate and produce disconnected crossover points. The braided sleeve was again carefully removed from the steel mandrel.

Both braided sleeves were manufactured with a diameter of 4.8 mm, a length of 50 mm and a braid thickness of 500 µm. The initial braid angle (α₀) for both samples was kept constant at 30°.

3.2.2. Fabrication of Braided Muscles

The muscle was fabricated in a similar fashion as that used to prepare the hydraulic Mckibben muscles introduced in Chapter 2. Inner tubes with thickness, length and diameter of 0.28 mm, 65 mm and 4.65 mm, respectively, were used to assemble the completed muscle. A finished example is shown in Figure 3.3.
3.2.3. Actuation Testing

Actuation testing was performed using the same set up as described in Chapter 2.

3.3. Results and Discussion

3.3.1. Actuation Testing of the McKibben Muscle Using the Braided Sleeve with Connected Junctions

The two different muscles prepared with and without strong connection at the crossover points were made and tested by pressurisation. The same amount of water pressure (0.66 bar) was injected into the muscles to compare the isometric and isotonic actuation behaviour. However, it was found that injecting the water into the M₁ muscle caused the braided sleeve to rupture immediately. Consequently, no strain or generation isometric force was measured for this sample. Bladder expansion in this sample would need to be accommodated by a braid deformation in which the fibres became bent between the crossover points, as illustrated in Figure 3.1b). The results show, however, that the printed PCL material was too brittle to
allow fibre bending and fibre rupture occurred. Consequently, the muscle with connected
crossover points failed to show any actuation movements and no further investigation was
performed on this type of material (Figure 3.4)

![Figure 3.4. The ruptured McKibben muscle after water injection.](image)

3.3.2. Actuation Testing of the McKibben Muscle Using the Braided Sleeve with Dis-
connected Junctions

The muscle constructed using the bladder made with dis-connected junctions could be
pressurized to 0.66 bar without failure. Figure 3.5 indicates that a maximum isometric
force of 960 mN was recorded for the 35 mm long muscle in just 1.3 s. Relaxation of the
generated force occurred completely on depressurization. These results were consistently
observed during four consecutive pressurization–depressurization cycles. This muscle
offers stress of 0.053 MPa (960 mN) which is 2.2 times less than HAM muscle introduced
in Chapter 2 with similar length and injected pressure (35 mm and 0.66 bar) but different
initial braid angle ($\alpha_0=30$ and $\alpha_0=35$). The performance of this muscle is also significantly
less than predicted data ($F= 5.96$ N, Contraction= 33.3%) obtained with equation 2.8. It is
most likely because of the different structure of the braided sleeve used in this Chapter. In this Chapter non-woven type of structure was used instead of woven structure which is common in conventional braided sleeves. Woven structure normally defines as a structure which the fibers are decussately on top and bottom of each other. In non-woven structure therefore the fibers are either on top or bottom of each other. Taking to account the diameter and length of this muscle, the generated isometric force is also reasonable in comparison to previous HAMs systems, as summarized in Table 1.4 (Chapter 1) [2-6].

Figure 3.5. Isometric force tests under constant water pressure (0.66 bar). Pressurization-depressurization test was performed four different cycles.
Figure 3.6. Isotonic actuation test under constant water pressure (0.66 bar) and given load of 12 mN; Pressurization-depressurization tests were performed for four different cycles.

An isotonic test was also performed under a load of 12 mN and water pressure of 0.66 bar applied for 1 s and then released. Figure 3.6 exhibits that the muscle contracted continuously during the pressurization period achieving a strain of 6.7 % in about 1 s. The strain fully recovered during depressurization and the muscle showed very consistent behavior during four pressurizing / depressurizing cycles. Similar to the stress generated, the contraction strain produced with this muscle is 1.7 times less than HAM muscle introduced in Chapter 2 with similar length and injected pressure (35 mm and 0.66 bar) but different initial braid angle ($\alpha_0$=30 and $\alpha_0$=35).
Figure 3.7. Typical static force and contraction strains emphasizing the role of input pressure and illustrating the hysteresis phenomenon for three different input pressures.

The static force verses contraction strain test was also performed for three different input pressures to investigate the hysteresis phenomena of this new muscle. Figure 3.7 exhibits that the amount of static force and contraction strain increase with increasing the input pressure similar to HAM systems introduced in Chapter 2. The maximum static force and contraction strain of 275 mN and 2.1% respectively, were achieved with input pressure of 0.45 bar. It wasn’t possible to perform the experiment for higher pressures as it caused some damage to the muscle. Only low pressure input of 0.66 bar was used to obtain isometric and isotonic diagrams in this chapter and this resulted in a smoother contraction and force generation output than reported data in Chapter 2 as a consequence of significantly less vibration produced by water pump.
The calculated power per mass indicates that, the muscle produce the maximum power per mass of 0.036 W/kg after just 0.85 seconds. The calculated power per mass declined once the injection of the water was stopped.

![Graph showing power per mass over time](image)

**Figure 3.8.** Corresponding power output from isotonic test; (0.66) and given load of 12 mN.

### 3.4. Conclusion

An alternative way to produce braided sleeve using 3D printing machine has been introduced for the first time. The effect of fibre connection in crossover points has been investigated. In this particular study it has been found that, the braided sleeves with connected fibres are unable to produce any actuation movements or isometric forces. The hydraulic McKibben muscle made of 3D printed braided sleeve exhibited 6.7% and 960 mN contraction strain and isometric force respectively with 0.66 bar injected water pressure.
3.5. References


Chapter Four

Thermally activated paraffin filled McKibben artificial muscles

This chapter presents the study that has appeared in the publication:

4.1. Introduction

As described briefly in Chapter 1, the need for heavy and bulky compressors/pumps makes pneumatic or hydraulic McKibben muscles unsuited to be employed as microactuators or in portable applications, where a highly compact design and weight minimization is required. The compressors/pumps are normally utilized to inject the pressurized fluids into the muscle [1-3]. The pressurized fluids are normally used to increase the volume of the inner bladder and subsequently deform the braided sleeve that make up the McKibben muscle[4]. The basic working concept of McKibben artificial muscles is that the braided sleeve translates the volumetric increase of the inner bladder to a lengthwise contraction of the braid that is capable of generating contractile forces much greater than an equivalent hydraulic or pneumatic system.

One approach towards making a more compact and lightweight actuation system is to reduce the need for compressors, pumps and valves by using a volume change material to deform the braided sleeve. Tondu and co-workers [5-7] have shown that pressurized gas/water can be replaced with pH sensitive hydrogel spheres in McKibben artificial muscles to generate reasonable actuation strain and force. However, there are still some remaining problems that need to be considered, such as the long response time (> 10 min), and the required pump for delivering acid/base solutions to the pH sensitive hydrogel. Recently, Sutter and co-workers[8] have also developed an enclosed system where enzyme catalysed hydrolysis of urea generates sufficient CO₂ gas to power a pneumatic McKibben muscle. In this chapter, we attempt to manufacture a novel McKibben artificial muscle filled with paraffin wax as an expandable temperature sensitive material. Paraffin wax has been shown to offer high thermal stability and gives volumetric expansion of ~20% when heated from 30 °C to 90 °C and ~10% extra expansion between 90 °C and 210 °C [9]. The volume change expected
during the full contraction of a McKibben muscle with starting braid angles of 30-40° and activated by pressurized fluid is of the order of 21%-78%. Therefore, it seems reasonable that the thermal expansion of paraffin within the McKibben muscle should be able to generate useful contraction strains and forces. Thermal expansion of paraffin has recently been used to create a new generation of artificial muscles by employing twisted/coiled carbon nanotube and niobium nanowire yarns, demonstrating successful torsional and linear actuation[9, 10].

The aim of the present chapter is to evaluate the performance of paraffin-filled McKibben muscles. Initially, paraffin wax-filled McKibben muscles were fabricated and heated using an external water bath. This system was used to evaluate the feasibility of using an expandable fill material to power the actuation of the McKibben muscle and to develop a quantitative model of output force and contraction strain for a given wax temperature. Secondly, a more practically useful wax-filled McKibben muscle was fabricated with an in-built electrical heating element. The output force, contraction strain and response time of both systems were evaluated.

4.2. Modelling of temperature driven McKibben artificial muscle

The most common approach[11] to model an ideal pressure-driven, cylindrical McKibben artificial muscle relates the static force \( F \) produced by the muscle to its contraction strain \( \varepsilon \) at various pressure differences \( P \) of the fluid contained inside the bladder within the braided sleeve compared with ambient pressure and as defined previously in Equation 1.4. This model assumes a full transmission of the pressure inside the inner bladder to the external braided sleeve; ignores ‘end effects’ relating to the non-cylindrical ends of the clamped braid; and does not include the effects of braid friction. To consider the impact of bladder stiffness on the muscle performance, theoretical[12] and semi-empirical [13] modifications have been added to the model, as described in Chapter 2.
All models originated from equation (1.4) treat pressure as an input variable to correlate generated force with strain. In the pneumatic and hydraulic McKibben muscles this applied pressure is easily measured and can be controlled as an input signal. In these conventional McKibben muscle systems, there is essentially an infinite reservoir of fluid available to maintain the desired pressure as the volume of the McKibben muscle changes. However, in a McKibben system operating by temperature-induced volume expansion of an inner fill material (e.g. paraffin in the current study), the volume change is finite and dependent upon the starting volume and pressure-dependent thermal expansion of the fill material. Also, the controlling parameter in these systems is the applied temperature $T$, so the model presented in equation (1.4) must be reformulated to replace $P$ with $T$. From braid geometry, the braided sleeve volume $V$ is directly related to the axial contraction strain by the following equation:

$$V(\varepsilon) = V_o \left[ b(1 - \varepsilon) - \frac{a}{3} (1 - \varepsilon)^3 \right]$$  \hspace{1cm} (4.1)

Here, $V_o$ is the initial volume within the braided sleeve, and $a$ and $b$ are the same as in equation (1.4). This equation suggests that the braided sleeve’s volume is known at any strain $\varepsilon$ with the braid’s geometry directly impacting this relationship through parameters $a$ and $b$.

Pressure will also affect the fill material volume and the coefficient of compressibility ($\kappa$) defines how pressure varies with material volume at a constant temperature:

$$\left( \frac{\partial P}{\partial V} \right)_T = -\frac{1}{\kappa V}$$  \hspace{1cm} (4.2)
Assuming $\kappa$ remains independent of $P$ and $V$ at low temperature and pressure ranges, from equation (4.2) pressure can be stated as a function of volume:

$$P = P_o + \frac{1}{\kappa} \ln(V_o/V) \quad (4.3)$$

$P_o$ in equation (4.3) is the starting pressure at which volume is $V_o$ and the temperature is $T_o$. Moreover, $\kappa$ is related to the thermal expansion of paraffin ($\alpha$) and thermal pressure coefficient of paraffin ($\gamma$) as:

$$\kappa = \frac{\alpha}{\gamma} \quad (4.4)$$

where $\gamma$ and $\alpha$ are, respectively, $(\partial P/\partial T)_V$ and $\frac{1}{V} (\partial V/\partial T)_P$. Both $\gamma$ and $\alpha$ can be determined experimentally by, respectively, measuring pressure as a function of temperature at a constant volume, and monitoring volume as a function of temperature at a constant pressure. Assuming both $\gamma$ and $\alpha$ are constant and independent of pressure, temperature and volume, equation (4.4) can be used to calculate $\kappa$ from experimentally measured $\gamma$ and $\alpha$. When $\kappa$ is known in equation (4.3), equations (4.1) and (4.3) are used to replace $P$ in equation (1.4)

$$F(T, \varepsilon) = (\pi r_o^2) \left[ \gamma(T - T_o) - \frac{1}{\kappa} \ln \left( b(1 - \varepsilon) - \frac{a}{3} (1 - \varepsilon)^3 \right) \right] [a(1 - \varepsilon)^2 - b] \quad (4.5)$$
where $T_o$ is the reference temperature at which $P = P_0$ and $V = V_0$.

Figure 4.1 schematically illustrates a McKibben muscle filled with a material that expands when heated and how the generated volume change and pressure deform the braided sleeve. Figure 4.6 shows the theoretical static force and contraction strain obtained from equation 4.5 that this muscle generates. In the unheated state the muscle is relaxed at reference point $O(P_0, V_0, T_o)$. Increasing temperature to $T_1$ in isometric mode (constant length) generates the maximum muscle force (or ‘blocked force’) at this temperature, as shown by state $A$ in Figure 4.1. By knowing how much pressure is generated in the muscle at state $A$, the blocked force can be obtained from equation (1.4) at $\varepsilon = 0$. From equation (4.1) it is seen that the isometric mode also corresponds ideally to a constant volume so that the pressure generated in the blocked state (i.e. state $A$) can be estimated to be $\gamma(T - T_o)$ when $P_0 = 0$ and $\gamma$ is independent of temperature. Under these circumstances, the blocked force for such muscle is calculated to be:

$$F_{block} = (\pi r_0^2)(T - T_o)(a - b)\gamma$$

(4.6)

By measuring blocked force at several temperatures, equation (4.6) can be used as a convenient way to obtain a value for the thermal pressure coefficient($\gamma$).
Figure 4.1. Schematic illustration of paraffin-filled McKibben muscle in starting (o), isometric (A) and isotonic (B) states indicating the relationship between experimental conditions and pressure, volume and temperature.

Figure 4.2. Schematic force verses strain diagram exhibiting different points plotted different pressure and volume at constant temperature.

The full performance envelope of an actuator system in terms of the mix of force and strain produced is illustrated in Figure 4.2. Experimentally, these data are collected by first measuring the blocked force under isometric conditions. Next the muscle is allowed to
contract to state $B$ and further until $F = 0$ while maintaining a constant input stimulus. The force / strain curve can also be determined theoretically and in the case of thermally-induced actuation of filled McKibben muscles, the behaviour is expected to follow equation (4.5).

4.3. Experimental

4.3.1. Paraffin filled McKibben artificial muscle fabrication

The paraffin-filled muscle for testing with an external water bath was fabricated as follows (Fig. 4.3). Firstly, a solid paraffin cylinder of 7.36 mm diameter was inserted into a thin latex rubber inner tube with a thickness and diameter of 0.28 mm and 7.40 mm, respectively. Next, the inner tube was inserted into the braided sleeve (polyphenylene sulfide (PPS), obtained from JDD TECH Company China) with a thickness of 0.44 mm and finally both ends of the muscle were sealed to prevent wax escape when the muscle was immersed in a water bath. The initial, unloaded length and diameter of the muscles were 35 and 8.8 mm, respectively. The initial angle ($\theta_0$) of the braided sleeve (Fig. 4.3) was determined by LEICA-M205 microscopy to be $34^\circ \pm 0.6^\circ$. 
The muscle with embedded heating element was also fabricated in a similar manner (Fig. 4.4). Firstly, a heating filament was inserted into the inner tube and then melted paraffin was poured into the inner tube. Once the paraffin set, the inner tube was inserted into the braided sleeve and finally the top and bottom of the muscle were sealed. The length, diameter and initial braid angle were 35 and 6.8 mm and $29^\circ \pm 0.9^\circ$, respectively. Although the same braid material was used to construct paraffin-filled muscles both with and without the embedded heating element, the method of fabrication resulted in slightly different braid angles and diameters.
4.3.2. Actuation test procedure

The experimental set up for actuation testing was specially designed to measure actuation strain, isometric force, response time and sample temperature. For the water bath tests, the actuation set up (Fig. 4.5) consisted of four main parts: a small hot plate, a small water container (80 ml), the paraffin filled McKibben muscle and a dual-mode lever arm force/distance transducer (Aurora Scientific, Model 300B). An e-corder data logger (ED 410,
e-DAQ) was also used to connect the lever arm unit to a computer, and e-DAQ Chart software was used to record the data. The temperature of water was monitored with a Digitech Qm-1600 thermometer. For the muscle containing the embedded heating element (Fig. 4.6), a DC power supply was used to control the voltage and current applied to the filament. The water bath was not used. An infrared camera (Micro – EPSILON/TIM160) was used to measure the surface temperature of the muscle.

Figure 4.5. Schematic illustrations of actuation set up of paraffin filled McKibben artificial muscle heated using an external water bath.

Figure 4.6. Schematic illustrations of actuation set up of paraffin filled McKibben artificial muscle heated using an embedded electrical heating element.
4.4. Results and discussion

4.4.1. Water bath heated paraffin filled McKibben muscle

The maximum forces generated by the paraffin-filled McKibben muscle were evaluated by immersing the muscle in a water bath, clamping the muscle ends to maintain a fixed length, and heating from ambient to five different bath temperatures ranging from 55 °C to 95 °C. The force-strain curves at each maximum temperature were also obtained by first allowing the muscle to contract in length and measuring the force at each contraction strain and then re-stretching the muscle to its original length. The obtained force/strain curves are shown in Figure 4.7. As expected, with increasing bath temperature the muscle produced higher blocked forces (at zero strain) and higher maximum strains (at zero force). The volume of the paraffin increases with temperature causing circumferential expansion of the braided sleeve and shortening of the actuator. Overall, the paraffin-filled McKibben muscle’s performance is very similar to that of the pneumatic or hydraulic McKibben muscles in which volume change of the braid is achieved by injecting pressurized fluid. However, the needed volume change to drive the paraffin-filled McKibben muscle occurs from inside the bladder without any connection to the outside world. The main limitation of these muscles, however, is the slow response time needed to heat the paraffin. The paraffin-filled McKibben muscle produced the highest static force and contraction free strain, 850 mN (or a stress of 17 kPa based on the muscle cross-sectional area) and 8.3%, respectively, at a bath temperature of 95 °C, which was the maximum temperature that could be reached. Melting tests confirmed that the paraffin used here began to melt at 55 °C. The lowest measurable blocked force and free contraction strain (95 mN and 2.5%, respectively) were produced at a bath temperature of 55 °C, or just on wax melting. No measurable actuation was detected at bath temperatures below
55°C, due to the small wax volume change at these temperatures. The static stiffness of the muscle prior to activation was high as the paraffin is solid in the dry state. All force/strain curves showed some hysteresis between the contraction and re-stretch cycles as is typical of conventional McKibben muscles and is likely related to braid friction [14]. All hysteresis curves show a reduced force for a given contraction strain on re-loading that during unloading.

Figure 4.7. Measured force and contraction strains produced by paraffin filled McKibben artificial muscle heated to different bath temperature, as indicated. All temperatures show same hysteresis direction.

Equation (4.6) was used to estimate the thermal pressure coefficient γ for the paraffin wax employed here. Using the blocked force data from Figure 4.7, the pressure at each maximum bath temperature was calculated from Equation (1.4) and these values were plotted against maximum bath temperature in Figure 4.8a. The calculated internal braid pressure exerted by the wax increased almost linearly with temperature and least-squares linear fit gave an estimate of γ of 87 Pa/K. The thermal expansion coefficient α for the paraffin wax (Fig. 4.8b)
was measured to be 0.0031 K\(^{-1}\) over the temperature range from 45\(^\circ\)C to 110\(^\circ\)C and this value is almost identical to that reported by Lima and co-workers [9]. The coefficient of compressibility \(\kappa\) was calculated as the ratio of \(\alpha\) and \(\gamma\) (equation (4.5)). Using these coefficients, equation (4.5) was then used to calculate static force \(F\) as a function of contraction strain \(\varepsilon\). The calculated results are shown as solid lines in Figure 4.9 and are in good agreement with the experimental data points. The good agreement between calculated and experimental values demonstrates the validity of the modelling approach based on the pressure-dependent thermal expansion of the fill material. The model has practical utility since desired muscle force and strain can be achieved by heating to the temperature given by equation 4.6.
Figure 4.8. (a) Change in pressure generated by heated wax within the McKibben muscles as a function of maximum temperature; (b) fractional volume change of paraffin wax as a function maximum temperature.

Figure 4.9. Typical static forces and contraction strains produced by paraffin filled McKibben artificial muscle with comparison to the model lines for each particular maximum temperature.
For this configuration of McKibben muscle where heating is provided through the water bath, the response time of the muscle is limited by the heating rate of the relatively large quantity of water (Fig. 4.10). The dynamic response of this system was evaluated during slow heating of the water bath (~8 °C/min). The maximum isometric force (730 mN) was generated after 8.5 minutes when the water bath temperature reached 95 °C. The rate of isometric force generation increases dramatically after 4 minutes of heating, as a result of melting and increased thermal expansion of the paraffin fill material. The isometric and isotonic cycle results were also fully reversible with a longer time needed for the returning cycles due to the slowness of the passive cooling of the paraffin wax and the surrounding water bath.

![Graph showing time variation of water bath temperature and corresponding isometric force produced by paraffin filled McKibben artificial muscle.](image)

**Figure 4.10.** Time variation of water bath temperature and corresponding isometric force produced by paraffin filled McKibben artificial muscle.

**4.4.2. Paraffin filled McKibben artificial muscle with heating filament**

The results of the previous section demonstrated that an expandable fill material, such as thermally-sensitive paraffin wax, can be used to power a McKibben muscle. However, the
water bath used for heating the wax is not a practically useful system, so a second set of samples were prepared that included an electrical heating filament embedded inside the wax. The force–contraction strain diagrams for these electrically heated paraffin filled McKibben muscles were obtained at six different voltage/current values applied to the filament (Fig. 4.11). The voltage ranged from 1.3 V to 5.8 V and current ranged from 0.30 A to 1.37 A. The muscle produced higher forces and strains with increasing the voltage/current as a consequence of higher temperatures generated within the wax. Input voltages greater than 5.8 V caused irreversible damage to the bladder and braid of the paraffin-filled muscle due to overheating at the connection to the heating filament. The muscle produced the highest static force and free contraction strain of, respectively, 2000 mN (71 kPa) and 9% at 5.8 V/1.37 A (7.94 W). The maximum force generated was well above that measured for the actuators heated in the water bath, indicating that much higher temperatures can be produced electrically than was practical to achieve with the water bath. Interestingly, the maximum contraction strain of the electrically heated muscle (9%) was similar to that generated at significantly lower temperature in the water bath (8%). The electrically heated muscles are significantly stiffer than those constructed without the heating filament, as indicated by the higher slopes of the force-strain curves shown in Figure 4.11 compared with the curves in Figure 4.7 The presence of the electrical heating element within the paraffin wax acts additionally as a mechanical reinforcement and increases the axial compressive stiffness of the system. As a consequence of this increased stiffness, the achievable contraction strain of this electrically heated McKibben muscle is restricted compared with the water-bath heated systems. Increased contraction strains of the electrically-heated system would be possible by developing a more compliant electrical heating filament.
In addition to enhanced practical utility, the electrically-heated paraffin filled McKibben muscles were expected to respond more quickly than the water-bath heated system in which response time was dictated by the large volume of water. Isometric tests were performed at different input voltage/current values and the force generated monitored with time (Figure 4.12a). The muscle reached 750 mN blocked force in just 1.5 min using 5.8 V/1.37 A (7.94 W). In comparison, the muscle heated in the water bath system reached the same isometric force after 8.5 min. It was also noted that the response time of the electrically heated muscle could be controlled by altering the supplied voltage/current. The muscle force reached a plateau after 12.5 min at the lowest voltage/current input (i.e. 2.3 V/0.55 A), indicating that temperature had reached steady-state equilibrium.
Figure 4.12. a) Isometric force verses time produced by paraffin filled McKibben artificial muscle-heating filament emphasizing the role of time on muscle performance and b) Surface temperature increases with time obtained with infrared camera.
The surface temperature of the muscle at each applied voltage/current was also measured using an infrared camera during the isometric tests (Fig. 4.13). The surface temperatures as a function of time (Fig. 8b) followed very similar trends to that of the force generation profiles (Fig 4.12). The isometric and isotonic cycle results were also fully reversible with higher response time for returning cycles. In the returning cycle the paraffin cooling process was highly dependent on convection heat transfer with surrounded environment.

Figure 4.13. Maximum surface temperature (within the white boxes) and images obtained with an infrared camera after 30 seconds of electrical heating for four different applied voltages/ currents (a) 2.3 V/0.55 C (b) 3.3 V/0.79 C (c) 4.3 V/1.02 C (d) 5.8 V/1.37 C. (The white squares indicate the approximate outline of the muscle).
4.5. Conclusions

A novel, compact McKibben type artificial muscle that utilizes an expandable fill material is introduced for the first time. Actuation is produced using a volume change of the fill material to increase the internal volume and cause simultaneous length contraction and tensile force generation. Thermally expanding paraffin wax heated electrically could generate a maximum force of 2N (71 kPa) or a maximum length contraction of 9%. The system does not require any pumps, valves or fluid tanks and is much more compact than a conventional fluid-driven McKibben artificial muscle. The wax-filled McKibben muscles were also characterized by controlled heating in a water bath. The experimentally produced forces and contraction strains were accurately predicted by the quantitative analysis developed here based on the input temperature and pressure-dependent thermal expansion of the paraffin. The muscles in both systems were cooled down naturally with surrounded air; as a consequence, the return response times were significantly longer than contraction response times.
4.6. References


Chapter Five

A bladder-free, non-fluidic, conductive McKibben artificial muscle operated electro-thermally

This chapter presents the study that has been submitted as a manuscript:

5.1. Introduction

As mentioned in the Chapter 1, McKibben artificial muscles that operate pneumatically or hydraulically provide excellent performance,[1, 2] very close to biological muscles in most regards[3], but require bulky pumps/compressors, valves and connecting lines. These devices add extra weight and volume to the actuation system and consequently increase the device packaging, which is a disadvantage for microactuator systems. Microactuator systems normally desire very compact and lightweight systems. As shown in Chapter 4, employing of a pressure generating material, such as thermally expanding paraffin wax, can eliminate the need for this additional infrastructure. The introduced artificial muscle systems in Chapter 4 offer promising and practical performance, specifically for microactuator systems. Operating without a pump/compressor, valves and tubing, the actuator system is significantly smaller and lighter in comparison to other fluidic McKibben muscles. The paraffin wax inside the inner bladder was directly stimulated by an embedded metal wire that is electrically heated. The paraffin thermally expands with sufficient volume change and pressure required to operate the muscle. The muscle generates an isometric force of 2 N (or 71 kPa stress) and 9% contraction strain after several minutes heating with a power supply of 7.94 W [4]. However, using the embedded heating element was found to restrict the muscle contraction as a consequence of metal element’s high stiffness. The embedded heating element also increases the system weight and limits the possibility of making smaller size muscle, which would theoretically offer significantly shorter response times as a result of faster heat transfer through the thermo-sensitive material.

The main aim of this Chapter is to further improve this concept by designing and developing a novel paraffin-filled McKibben muscle without the embedded heating element and the inner bladder to achieve a smaller size, lower weight muscle and a faster response. Through this
work it was found that incorporation of electrically conductive wires in the braided sleeve allows for convenient Joule heating of the paraffin, which makes the muscle independent of heating element or water bath.

This Chapter introduces a conductive and bladderless paraffin filled McKibben muscle by using a conductive braided sleeve with an optimized cover factor. As described in Chapter 2, the elastic expansion of the bladder is known to reduce the pressure available to work against the braid in a McKibben muscle, thereby reducing the muscle performance\cite{5}. The possibility of developing bladderless McKibben muscles was inspired by the recent demonstration of torsional and tensile actuation in paraffin-filled carbon nanotube twisted yarns \cite{6}. In these systems, the paraffin wax was successfully contained within the porous carbon nanotube yarn by surface tension during heating and cooling through the melting transition.

5.1.1. The effect of braided sleeve structure on performance of novel conductive and bladderless paraffin filled McKibben muscle

To eliminate the use of bladder in the new bladderless, paraffin filled McKibben muscle, two important concepts have been considered which are directly related to the braid structure. Proper braid analysis and design is needed to successfully prevent the paraffin leaking from the conductive braid even above the wax melting temperature. The suitability of containing molten paraffin within a braid can be evaluated using the approach used for porous membranes. The pressure needed to push a non-wetting liquid through the pores of a membrane is called the breakthrough pressure, $P$, and is related to the membrane and liquid properties by the following Young–Laplace equation \cite{7}:

$$P = -\frac{2\sigma \cos \theta}{r}$$  \hspace{1cm} (5.1)
Where, \( r \), is radius of the pores, \( \sigma \) and \( \Theta \) are the surface tension of the liquid and the contact angle, respectively. For any pair of materials, the breakthrough pressure increase as the size of pores decreases.

Pore sizes in a braid can be expressed in terms of the cover factor, \( C \), which is defined as the ratio of the area occupied by the yarn within a periodic pore unit to the total area of the pore unit [8], as shown in Figure 5.1. According to equation 5.2 the cover factor is a function of braid diameter, \( d_b \), initial braid angle, \( \frac{\alpha}{2} \), yarn width, \( w_y \) and number of threads, \( N_c \). In this research, the cover factor of the braided sleeve was varied and assessed in terms of its ability to prevent the paraffin wax exuding from the braided sleeve in the expanded state. The cover factor was varied by independently decreasing the diameter of the braid as well as increasing the yarn width.

\[
C = \frac{w_yN_c}{\pi d_b \cos \frac{\alpha}{2}} - \left[ \frac{w_yN_c}{2\pi d_b \cos \frac{\alpha}{2}} \right]^2 \quad 0 < \alpha < 180 \quad (5.2)
\]

Figure 5.1. The schematic view of conductive braided sleeve indicating the diamond shaped periodic pore unit. The right hand side schematic image includes four threads, or \( N_c=4 \). The width of the yarn \( (w_y) \) and the braid angle \( (\alpha) \) are also shown.
5.2. Experimental

The bladderless, conductive McKibben artificial muscles were fabricated as illustrated in Figure 5.2. Firstly, conductive braided sleeves were made with a braiding machine (Trenz-Export Apartado 133) using steel wires (0.035 mm diameter) and cotton fibers (0.143 mm diameter). The steel wire and cotton fibers were purchased from Shijiazhuang Yunchong Trading Co., Ltd and were prepared for braiding as a feed yarn consisting of one cotton fiber and one steel wire in parallel. Three different braids were constructed to vary the cover factor (Table 5.1). Braids of 1.4 mm diameter (1.6 mm after paraffin injection) were prepared using either a single feed yarn (K1) or two feed yarns in parallel (K2) to adjust the yarn width and cover factor. A third braid (K3) was made using double feed yarns, but to a braid diameter of 2.2 mm. The initial, unloaded length of all muscles was 40 mm. The initial angles (α/2) of the braided sleeves were determined using a LEICA-M205 microscope to be 34° (40° after paraffin injection) for K1 and K2 muscles and 44° K3 muscle (Figure 5.3 a, c and e).

Table 5.1. Comparison of three different conductive and bladderless McKibben artificial muscles.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Braid Cover factor</th>
<th>Braid Diameter (mm)</th>
<th>Initial Braid angle (°)</th>
<th>Yarn diameter (mm)</th>
<th>Molten Wax Contained within braid when heated above wax melting temperature?</th>
</tr>
</thead>
<tbody>
<tr>
<td>K1</td>
<td>0.73</td>
<td>1.4</td>
<td>34</td>
<td>0.286</td>
<td>Yes</td>
</tr>
<tr>
<td>K2</td>
<td>0.56</td>
<td>1</td>
<td>34</td>
<td>0.143</td>
<td>No</td>
</tr>
<tr>
<td>K3</td>
<td>0.63</td>
<td>2</td>
<td>44</td>
<td>0.286</td>
<td>No</td>
</tr>
</tbody>
</table>
The molten paraffin wax (Temperature: 90 °C) was injected into the braided sleeve using a fine needle with diameter of 0.7 mm (Figure 5.3b, d and f) and allowed to solidify by cooling. Finally, the top and bottom of the braided sleeve were sealed with rapid glue. Optical microscopy was used to determine whether the paraffin wax was contained within the braided sleeve after a five isotonic test of heating (T: 124 °C) and cooling cycles with 7.6 kPa given load. Only the K1 braid with the highest cover factor was able to prevent the wax exuding through the pores when heated (Table 5.1) and Figure 5.3. The K1 muscle was used for further actuator evaluation. The resistance and weight of the K1 sample were measured to be ~18 Ω and 0.14 g, respectively.
Figure 5.2. The schematic illustration of (a) braiding machine and (b) paraffin injection process into the braided sleeve. (c) Photograph of the entire muscle with connected wires.
Figure 5.3. Microscopy images of $K_1$ (a,b) $K_2$ (c,d) and $K_3$ (e,f) before and after paraffin (green colour) injection.
The actuation testing system (Figure 4) using a force-distance transducer (Lever Arm 300B, Aurora Scientific) was used to measure isotonic actuation strain, isometric force and response time. An E-corder data logger (ED 410, e-DAQ) was used to connect the lever arm unit to a computer, and e-DAQ Chart software was used to record the data. A DC power supply was also utilized to control the voltage and current applied to the braided sleeve for electrical heating. An infrared camera (Micro – EPSILON/TIM160) was also used to measure the surface temperature of the muscle.

![Lever arm and data acquisition](image)

Figure 5.4. Schematic illustration of actuation set up of bladderless, conductive McKibben artificial muscle connected to the voltage supplier.

### 5.3. Results and Discussion

Analyses of isotonic contraction and the response time of the conductive McKibben muscle were performed by stimulating the muscle with 2.5 volts (0.35 W) under six different constant stresses ranging from 7.64-127 kPa. The stimulation was discontinued once the
muscle reached the maximum contraction strain which was approximately 60 seconds in all cases. Figure 5.5 indicates that the muscle contracted continuously during the heating stimulation period and a strain as high as 10% was achieved at the smallest applied load (7.64 kPa). The maximum contraction strain decreased with increasing isotonic applied load and the muscle exhibited just 2.1% contraction strain in 60 sec under 127 kPa stress. The decrease in contraction strain with increasing applied load is likely due to a decrease in muscle structural stiffness[9] in the heated state as compared to the initial cooled condition. The work density was calculated from the maximum contraction strain at each applied stress and is shown in Figure 5.6. The maximum work density of 3.5 kJ/m\(^3\) was achieved under the constant load of 98 kPa, and is 43.75 % of the work density of natural muscle (8 kJ/m\(^3\)).

Figure 5.5. Dynamic behaviour of the conductive and bladderless McKibben muscle: isotonic contraction test under six different stress and constant voltage of 2.5 V.
Figure 5.6. Corresponding work density output calculated from the maximum contraction strain peak for each individual stress.

The cycle behavior of the conductive and bladderless McKibben muscle was investigated for five consecutive heat/cool cycles at an isotonic load of 127 kPa. It was observed that the muscle was able to expand approximately 0.5% strain which is 21% of its initial contraction strain during the cooling process as shown in Figure 5.7. After this first heat/cool cycle (165 sec), the muscle showed a very consistent behavior for the next four cycles with a completely reversible contraction and expansion occurring during heating and cooling, respectively. The average range of actuation contraction strain and expansion strain in heating/cooling cycles of Figure 5.7 is 0.39% ±0.08 and 0.355% ±0.065, respectively. Friction between braid fibers may have restricted the amount of expansion occurring in the initial cooling cycle. Once the muscle was frozen to a different length, diameter and initial braid angle, the amount of contraction strain in the following cycles was then diminished [9]. In a second set of experiments the sample was manually stretched to its initial length immediately after stopping the heating stimulation. It was found that the muscle exhibited the same amount of large contraction strain during each subsequent heating process for
three consecutive heating and cooling cycles (Figure 5.8).Resetting the muscle to its starting dimensions after each heating cycle allowed the full contraction strain to be developed in the subsequent heating process.

Figure 5.7. Contraction strain verses time for five different cycles under 127 kPa load without external re-stretching.
Figure 5.8. Contraction strain verses time for three different cycles under 127 kPa load with external stretching during the cooling process. (Dashed lines represent the manually stretching).

Isometric tests were also performed at three different input voltages/currents (Figure 5.9) to investigate the ability of muscle to generate force. The muscle was able to generate up to 39 mN isometric force (50 kPa stress) in just 20 sec. The isometric force showed a very consistent cycle behavior with fully reversible force generation and relaxation during heating and cooling, respectively. The length and diameter of the braided sleeve were constant during these experiments, unlike in the isotonic tests, which accounts for the consistent force generation during consecutive cycles. Figure 5.10 indicates that the maximum surface temperature of 125 °C was achieved after 20 sec when an electrical power of 0.35 Watt was applied. According to previous work, the expected maximum force of the paraffin-filled McKibben muscle can be calculated using equation 4.6 [4].
At the measured maximum temperature, the calculated maximum force is 34 mN, which is very close to the measured force generated (35-39 mN).

Figure 5.9. Dynamic behaviour of the conductive and bladderless McKibben muscle: Isometric force verses time for three different voltages/currents and three continues cycles.
Figure 5.10. Surface temperature images obtained with an infrared camera for an input voltage of 2.5 V and current of 0.14 A during the isometric test. The dark blue colour always represents the lowest temperature (22.6 °C) and the yellow colour represent 35.0 °C, 63.9 °C, 100.3 °C, 124.9 °C for 5, 10, 15, 20 seconds, respectively. (Bottom areas indicate higher temperatures as a result of being closer to the electrical connections).

This bladderless McKibben muscle offers almost the same amount of contraction strain (10%) and stress (50 kPa) as the previously reported paraffin-filled McKibben artificial muscle with embedded heating filament (9% strain and 71 kPa stress) [4]. The smaller diameter of the bladderless McKibben muscle means that it responds considerably faster (20 seconds) than the previous system (90 seconds) where a larger diameter was needed to accommodate the embedded heating element. The bladderless system also used 23 times less power (0.35 W vs 7.94 W) to reach the peak force and strain and 4 times less power to generate the same amount of isometric force as the previously described system.
Figure 5.11. Typical static forces and contraction strains produced by bladderless McKibben muscle emphasizing the role of applied voltage and illustrating the hysteresis phenomenon.

The force-strain curves for two different applied voltages were also obtained by first allowing the muscle to contract in length and measuring the force at each contraction strain and then re-stretching the muscle to its original length. The obtained force/strain curves are shown in Figure 5.11. As expected, with increasing applied voltage the muscle produced higher blocked forces (at zero strain) and higher maximum strains (at zero force). The volume of the paraffin increases with temperature causing circumferential expansion of the braided sleeve and shortening of the actuator. Overall, the bladderless McKibben muscle’s performance is very similar to that of the pneumatic or hydraulic McKibben muscles in which volume change of the braid is achieved by injecting pressurized fluid. However, the needed volume change to drive the bladderless McKibben muscle occurs from inside the conductive braid without any connection to the outside world. The bladderless McKibben muscle produced the highest static force and contraction free strain, 50 mN and 4.8%, respectively, for applied
voltage of 2.5 V. The blocked force differences in loading and unloading curves for the same applied voltage is likely because of temperature increase during this process, which leads to production of higher forces in loading curves. Interestingly, the same phenomena have been observed for both applied voltages with the same blocked force difference of 20 mN. The lowest measurable blocked force and free contraction strain (16 mN and 3.2%, respectively) were produced for applied voltage of 2.2 V. Both curves exhibited a similar degree of hysteresis in the loading and unloading curves that has been attributed previously to braid friction[10].

5.4. Conclusions

A conductive and bladderless McKibben artificial muscle is introduced for the first time. The conductive braided sleeve was made of intertwined steel wire and cotton fiber with a diameter of 1.4 mm. The temperature sensitive material (paraffin) was successfully constrained inside the conductive braided sleeve even at expanded state by increasing the yarn width and adjusting the braid angle to give a high cover factor of 0.89. The muscle generates a maximum tensile stress of 50 kPa and maximum contraction strain of 10% in 20 and 60 sec, respectively, with a small input voltage of 2.5V. Using this design, electrically-powered and small diameter McKibben muscles can be developed for micro-actuator applications.
5.5. References


Chapter Six

Conclusion and Future work
6.1. Summary and conclusion

Pneumatic McKibben artificial muscles suffer from several disadvantages, such as: a noisy system due to exhaust during depressurization, heavy and bulky actuation system to carry for human or robots and high electricity consumption [1, 2]. This type of McKibben muscle is therefore unsuitable where compact size and weight minimization are required [3]. The aim of this presented thesis was to develop new types of McKibben artificial muscles by eliminating the current disadvantages and subsequently introduce a suitable artificial muscle system, which can be used in microactuator systems. Microactuator systems normally require very compact and light artificial muscles.

Recently, a series of studies have been undertaken to introduce new types of McKibben artificial muscles by replacing the conventional fluid (gas) used in pneumatic version of McKibben muscles with water, oil or pH sensitive hydrogel spheres [4-9]. The pressurized liquid or pH sensitive hydrogel spheres function similarly to the gas by increasing the volume of the inner bladder and subsequently stimulate the braided sleeve. As a result, the muscle either shrinks or expands in the length direction depending on the initial angle of the braided sleeve. Indeed, new types of McKibben muscles introduced in these studies are as practical as pneumatic version of McKibben muscles. However, the pH sensitive version still suffers from high device packaging as a consequence of using pump and pipes as well as long response time. The response time of these muscles is about 10 minutes which is 600 times higher than that of pneumatic version of McKibben muscles. The hydraulic version of McKibben muscles can be used in a fully enclosed system, which is useful for robotic devices, but pumps and pipes are still required in these systems.

The main purpose of using inner bladder in manufacturing pneumatic, hydraulic and pH sensitive hydrogel versions of McKibben artificial muscles is to keep the fluid or sensitive
material inside the braided sleeve in rest and fully inflated states. Recently, Meller et al. [8] and Pillsbury et al.[10] have shown that the effect of using inner bladder on pneumatic and hydraulic versions of McKibben artificial muscles. It was found that the inner bladder reduced the pressure available to work against the braided sleeve and consequently restricted the McKibben muscle output. A semiempirical model was also introduced by Meller et al.[8], which is able to accurately predict artificial muscle performance by taking to account fitting parameters obtained from experimental data. The classical model introduced by Tondu and Lopez [11] assumes full transmission of the pressurized stress inside the inner bladder to the external braided sleeve. Therefore, this model is unable to consider the bladder stiffness and is therefore inaccurate.

The main aim of Chapter 2 [1] of this thesis was to investigate the effect of bladder stiffness and muscle geometry on a small hydraulic McKibben muscle as well as the possibility of running this system with a low voltage water pump. An acceptably accurate model was also introduced to predict the effect of muscle performance. The new model takes two important parameters into account: the required pressure to inflate the inner bladder to make contact with braided sleeve \(P_{th}\), and the pressure needed to elastically deform the inner bladder in order to stimulate the braided sleeve. As detailed/discussed in chapter 2, hydraulic McKibben muscles were scaled down to a diameter of 6 mm and lengths of 35–80 mm. Isometric force and isotonic strain of 26 N and 23% were obtained respectively. The overall system response time was 1.1 s.

The braided sleeve is also an important segment of McKibben artificial muscles and is normally made using industrial braiding machines. However, there are several disadvantages in manufacturing the braided sleeves with commercially available braiding machines. In Chapter 3 of this thesis we investigated 3D printing as an alternative method to manufacture
braided sleeves. Two different types of braided sleeves with connected and disconnected fibers in junction points were printed. The effects of this connection on the performance of McKibben muscle were studied. It was found that the braided sleeves with disconnected fibers were more practical in the manufacturing of McKibben artificial muscles. The 3D printing method was faster and offered more accurate tools in manufacturing of these braids in comparison to conventional braiding machines. A hydraulic McKibben artificial muscle with a diameter of 4.8 mm and length of 35 mm was also assembled using disconnected printed braid. The muscle produced 6.7 % contraction strain and 960 mN isometric force, with 0.66 bar injected water pressure.

The principal problem investigated in this thesis was the introduction of a new McKibben muscle system which can operate without the need of compressors, pumps or piping. The first step in pursuing this ambition was to find a material that could be used instead of air or water inside the inner bladder. The material that could potentially exhibit volume change without any need to be connected to out of the inner bladder as the only reason of using compressor/water and piping in conventional versions is to deliver the fluid (into the inner bladder) to increase the volume of the inner balder. Paraffin wax, as a temperature sensitive material, has been used by other research groups to drive artificial muscles [12, 13]. Paraffin is a thermally stable material, which increases in volume by 20% when heated from 30 to 90 °C. This amount of volume expansion is sufficient to drive a Mckibben muscle with initial braid angles of less than 40°. The required temperature range in this case is also feasible to provide by applying low voltages. Thus, it seems possible to replace the air/water with paraffin due to its ability in increasing the volume of the inner bladder. Interestingly, paraffin can offer volume increase without requiring more material from the outside of the inner bladder. This property is crucial in manufacturing a Mckibben muscle system with no
pumps/compressors or piping. Therefore, in this thesis, the possibility of manufacturing paraffin drive McKibben muscle was also investigated.

In Chapter 4 [3] of this study, the air/water/pH sensitive spheres used in previous types of McKibben muscles were successfully substituted with paraffin wax. Two different types of paraffin driven McKibben muscles were introduced in this chapter. The first muscle was immersed into a water bath and the temperature of the bath was gradually increased to 95 °C from room temperature to stimulate the paraffin wax inside the muscle. The muscle produced the maximum isometric force of 850 mN and the maximum contraction strain of 8.3% at the highest possible provided temperature (95 °C). The second paraffin driven muscle was stimulated through an embedded heating filament inside the inner bladder. The voltage/current was increased from 1.3V/0.30C to 5.8V/1.37C to stimulate this muscle. The muscle produced 2 N of isometric force and 9% contraction strain. The response time of this muscle was lower compared to the first muscle as a result of the different heating methods. Both muscles showed reversible actuation movements, with longer response time in the return section. The need to use a compressor, pump or piping was successfully eliminated in both muscles. A quantitative model, with a reasonable accuracy, was also introduced to predict the force and strain outputs of thses muscles at different input temperatures.

As mentioned earlier, the inner bladder used in the conventional version of McKibben muscles consumes some of the provided pressure inside the muscle before transmitting it to the braided sleeve. This phenomenon affects the performance of the muscle by reducing the expected isometric force and contraction strain. Chapter 5 of this thesis investigated the possibility of manufacturing bladder-free McKibben artificial muscles for the first time. In this chapter the possibility of removing the heating filament used in the previous muscle was studied. The conductive braided sleeve was made of intertwined steel wire and cotton with a
diameter of 1.4 mm. The paraffin wax was successfully kept inside the conductive braid, even in a molten state, by choosing proper braid cover factor (0.89). The muscle produced a maximum isometric force of 39 mN and maximum contraction strain of 10% at applied voltage of 2.5 V. The response time of this muscle was significantly lower than previous systems introduced in Chapter 4 as a consequence of the smaller size and therefore faster heat transfer. The performance of this muscle was also accurately predicted with the equation introduced in Chapter 4. This muscle can be developed to be used in microactuator systems.

Table 6.1. Properties of paraffin filled McKibben muscles compared to the common contractile artificial muscles and biological muscle.

<table>
<thead>
<tr>
<th>Muscle type</th>
<th>Blocking stress (kPa)</th>
<th>Free strain (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Contractile biological muscle[14]</td>
<td>100-350</td>
<td>20-40</td>
</tr>
<tr>
<td>Shape memory alloys (NiTi)[15]</td>
<td>200000</td>
<td>10</td>
</tr>
<tr>
<td>Artificial muscles from fishing Line and sewing thread[16]</td>
<td>22000-140000</td>
<td>49</td>
</tr>
<tr>
<td>Paraffin filled McKibben muscle for using in external water bath (Chapter 4)</td>
<td>17</td>
<td>8.3</td>
</tr>
<tr>
<td>Paraffin filled McKibben muscle with embedded heating element (Chapter 4)</td>
<td>71</td>
<td>9</td>
</tr>
<tr>
<td>Bladder-free, wax filled McKibben artificial muscle (Chapter 5)</td>
<td>50</td>
<td>10</td>
</tr>
</tbody>
</table>

Table 6.1 indicates that the filled McKibben artificial muscles introduced in this thesis generate significantly less blocking stress in comparison to biological muscles and common contractile type of artificial muscles such as shape memory alloys and fishing line artificial muscle. The bladder-free wax filled McKibben muscle offers 10% free strain which is similar
to shape memory alloys and half of the amount of free strain offer by biological muscles. The free strain offer by bladder-free wax filled McKibben muscle is well below that 49% produced by fishing line muscle. The non-linear behavior of the wax filled McKibben muscles and hysteresis are also issues that require further research. Performing actuation tests in a completely sealed system to prevent nonlinear heat transfer may potentially result in more linear behavior of these muscles.

6.2. Future work

The 3D printed braided sleeve introduced in Chapter 3 was manufactured using only one type of polymeric material (PCL). Further work is required to investigate the possibility of using different types of polymeric materials for comparison purposes. Furthermore, only 10 different individual fibers were used in manufacturing the 3D printed braided sleeve. As mentioned earlier, the cover factor of the printed braided sleeves could be increased by increasing the number of fibers used in manufacturing the braid. Low cover factor in this case limited the injected water pressure to 0.66 bar only as the inner bladder came out of the braid in higher injected water pressures. Additional experimental work could be performed to increase the cover factor of these braids and consequently open up the possibility of operating these muscles in higher injected pressures.

In Chapter 5 of this thesis, the conductive braided sleeve was fabricated from steel wires and cotton fibers by a conventional braiding machine. Further work could be done to increase the conductivity of these braids by using more conductive metals such as silver, copper and gold. More conductivity would potentially lead to lower response time and higher efficiency of these types of McKibben muscles. Actuation reversibility of these muscles could be also improved by using different types of fibers instead of cotton to ultimately reduce the friction
between fibers as well as fibers and steel wires. The inclusion of temperature sensors in the braided sleeve would allow for better control of the force and strain output.

6.3. References