Electromagnetically Responsive Soft-Flexible Robots and Sensors for Biomedical Applications and Impending Challenges

Hritwick Banerjee and Hongliang Ren

Abstract Advantages of flexible polymer materials with developments in refined magnetic actuation can be intertwined for a promising platform to work on a resilient, adaptable manipulator aimed at a range of biomedical applications. Moreover, soft magnetic material has an inherent property of high remanence like the permanent magnets which can be further refined to meet ever-increasing demands in untethered and safe-regulated medical environments. In this chapter, we focus mostly on different avenues and facets of flexible polymer materials in adaptable actuation and sensing in the context of magnetic field for range of biomedical applications.

1 Introduction

Instead of rigid, inflexible, stiff robots, the era of robotics evolving rapidly towards a soft, flexible, yet resilient and squishy counter parts. The advantage of using a soft material as active component comes with its own challenges. In theory, soft materials will pose infinite DOF, which makes control, and design an increasingly difficult problem for engineers. Therefore, we need to have a better trade-off between a combination of rigid and soft counterparts to support objective optimally. Electromagnetic actuation is widely been used to handle the movement of medical robots for decades now in the realm of biomedical field for its wireless, untethered and safe control. Along with this, soft magnetic material has an inherent property of high remanence like the permanent magnets to be used as a better regulated control [35, 39, 70]. Due to the non-linear relationship between electromagnetic torque and bending angle of the soft material, quantization of the magnetic field inside a deformable structure is a nontrivial problem to investigate [1]. In the paradigm of smart materials, magnetic fields and its field distribution have been studied, tested and used in biomedical systems such as magnetic sensors, magnetic nanoparticles for MRI and multifunctional drug releasing polymers [69]. In this chapter, we will focus on the fabrication and

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development of flexible polymer materials in an adaptable manipulator which will be magnetically actuated [52, 97].

1.1 From Magnetorheological (MR) Fluid to Magnetorheological Elastomer

The magnetorheological fluid is kind of a smart and controllable materials by mixing ferromagnetic particles in oil or water [40]. Under the magnetic field, the viscosity of the fluid changes and it becomes a viscoelastic solid. The MR fluid is affected by the difference between an elastic modulus with applied magnetics fields (on-state) and the same modulus with no field applied (off-state). The factors affect the MR effect includes the matrix's elastic modulus, the magnetic properties of the particles, size and concentration of the conductive particles [47].

To understand more in detail of the MR fluid, there is an inherent necessity to understand the inherent mechanics of the materials with magnetic micro particle (mostly in 1–10 micron range) (Table 1).

1.2 MRF Mechanical Dynamic Behaviour

When there is no electromagnetic field applied to the MR fluid, this behaves almost similar to the Newtonian fluids. Therefore, a simple Bingham model is well suitable to describe the dynamic behaviour of the field-dependent fluid characteristics. On the contrary, in the context of non-Newtonian fluids, Bingham plastic model will behave such as to increase the yield stress before there is any sort of flow imparted [16]. Along with the current scenario, there are numerous models been presently studied as referenced herein [12, 100]. Along with the line of Bingham model, the shear stress strain curve follows the behaviour as described [40].

Table 1 Summary of the properties of MR fluids (Data depicted from [40, 58]).

Property	Typical value
Initial viscosity	0.2–0.3 [Pa s] (at 250C)
Density	3–4 [g/cm^3]
Magnetic field strength	150–250 [kA/m]
Yield point	50–100 [kPa]
Reaction time	Few milliseconds
Typical supply voltage and current intensity	2–25 V, 1–2 A
Working temperature	−500 to 1500 °C

1.3 Magnetorheological Elastomer [MRE]

MR Elastomer consists of polymer media like silicon or natural rubber with materials that can be polarized. MRE exhibit similar characteristics to that of MR fluid but the MR fluid operates at post yield region whereas MR elastomer operates at pre yield. The major shortcomings faced by MR fluids like the contamination, deposition of iron particles are overcome by MRE [40]. There are various classification of MRE based on the distribution of particles, structure, electrical and magnetic properties (Fig. 1).

During curing or crosslinking of matrix, magnetic field is applied to the polymer composite. This would lock the particles in a columnar chain structure making it anisotropic [58]. When the field is not applied, the iron particles are randomly oriented making it isotropic. The choice of curing the elastomer composite is purely dependent on the type of application. Due to this columnar structure in anisotropic material, there is a low dipolar energy state. Shear modulus and the work required to displace the particle from the low energy state will require higher magnetic field.

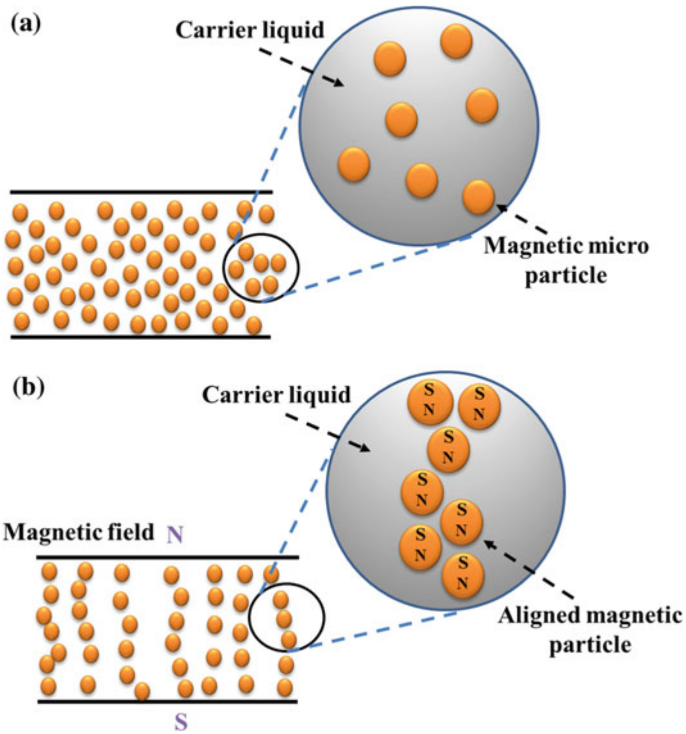

Fig. 1 MR fluid with and without magnetic field. **a** Without an external magnetic field **b** With introduction of an external magnetic field to align the magnetic micro particles.

1.4 Magnetically Responsive Stiffness Tuning in Flexible Manipulators

The advantages of flexible polymer materials with advances in refined magnetic actuation can be combined for a promising platform to work on a resilient, adaptable manipulator for a range of biomedical applications. In this realm, there are published report where microelectro-mechanical systems (MEMS) are embedded in magneto rheological polymer for different motion control [42]. In the regime of magnetorheological elastomer (MRE), experiments shown to have optimal magnetic effect with 60% carbonyl iron particles content mixed in an anisotropic solution of silicon rubber and silicon oil mixture [30]. To further optimize the whole magnetic system, researchers came up with a particle model with the help of finite element model (FEM) analysis to counteract the conjunction of ferromagnetism with viscoelasticity [60].

2 MRE for Biomedicine

MREs have the actuation properties close to that of natural muscles while the same principle can be extrapolated for peristaltic devices like micropumps. Apart from artificial muscles, MREs can also be applied widely in drug delivery as the principle mechanisms of TDD devices also based on fluid pump functioning. In the realm of drug delivery, recent advances mostly concentrated in controlled drug administration where magnetic field can take a promising action [47]. So to understand and administer in depth MRE controlled action, it is needed to appreciate the mechanical system more aptly as briefly described herein (Fig. 2).

2.1 Properties of MRE

MRE properties strongly depends on the magnetic field strength. There is a change in shear modulus and stiffness when the MRE materials are deformed [72]. Their response rate is usually in order of millisecond.

Properties of MRE materials depend on various factors.

(1) The shear or elastic modulus, density is due to the material properties of the elastomer matrix
(2) The magnetizable property depends on the micron-sized particles selected.

Hence, it is highly important in choosing the materials for MRE. The type of material used for the elastomer matrix will highly determine the viscoelasticity of MRE. Using ferromagnetic particles of size $<1.5\ \mu m$ and assumed to magnetize uniformly. In these materials, the maximum change in modulus or stress for maximum

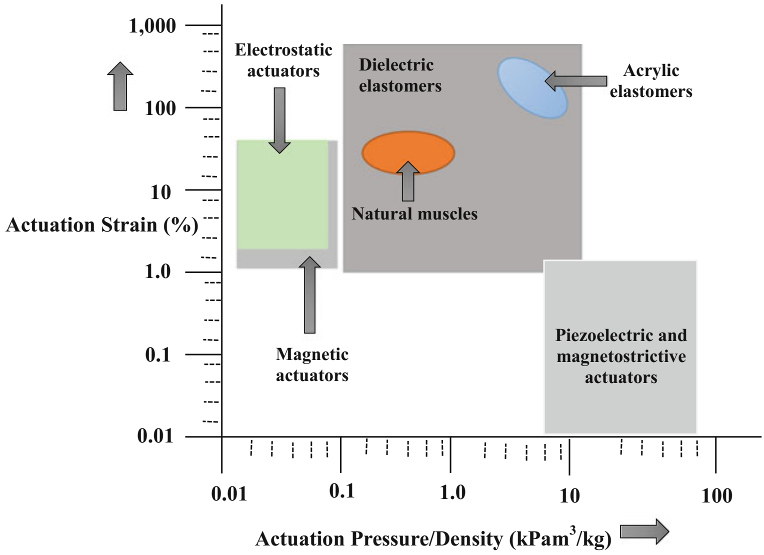

Fig. 2 Different regime of soft actuators and natural muscle [4].

magnetic field applied occurs when the magnetic particles in the MRE are saturated. The pure iron and alloys of iron exhibit this property and cobalt as these materials has high magnetic saturation. In most of these MRE materials, modulus has a significant increase when the stress applied is parallel to the magnetic particle in the elastomer and also anisotropic MRE materials exhibit a higher increase in modulus than randomly oriented magnetic particle.

The main factor that determines the property of MRE is the type of the ferromagnetic materials used like the percentage volume, type of distribution, the shape and size of the particle. It is observed that there is a low core loss and eddy current loss within the MRE.

2.2 Change in Property Under Magnetic Field

Properties of MRE materials like elasticity, plasticity and viscosity depends on the applied magnetic field. This study of deformation of the material under the magnetic field is required for its application. These variables can be controlled by the use of changing magnetic field. According to research result, for applied magnetic field of 0–0.3T, the range of shear strength of the material varies from 2–3 to 100kPa. There is also an influence on the composite properties due to the alignment of magnetic particles with or without external magnetic field [46]. For both anisotropic and isotropic MREs, the material properties like the stiffness and damping can be altered using an external field. With the applied magnetic field and optimizing the alignment of particle and density, the damping or the stiffness can be increased [8]. For isotropic material, the increase in stiffness or damping can be observed in applied field if the volume fraction of the filler like the iron exceeds 15% [8]. For typical elastomer, the shear modulus increase due to the magnetic forces of the inter-particle is almost 50% to the zero field shear modulus. Hence from this, the shear modulus of elastomer which are cured in the presence of magnetic field, i.e. anisotropic is no larger than the shear modulus of the same elastomer with randomly distributed particles, i.e. isotropic [99]. From various research, 27% of volume fraction of iron particles was predicted to be optimum value [36, 98].

2.3 Flexibility and Tunability

Susceptibility (or permeability) of different types of magnetic elastomers (e.g. difference between magnetic properties of elastomers with magnetic particles, phase composition) changes greatly with the electric field potential. For example, in AC magnetic fields, iron particles are continually conductive and hence heating of the devices increased up dramatically. This was due to the heating via absorption of AC magnetic field, MRE resulting in hyperthermia processes. On the contrary, high-intensity DC magnetic fields lead to hard magnetic filler, which further results in changing anisotropic properties and overall matrix elasticity.

Carbonyl iron micro particles have the highest possible saturation magnetization and high particles loading which is established up to 75 wt % with respect to polymer [28]. The relationship between saturation magnetization is that it increased linearly with particle concentration increasing. The soft magnetic and soft mechanic elastomer with 50 wt % Fe exhibited 90 emu/g saturation magnetization and 80% elongation in magnetic field of 1 T and stronger.

2.4 Sensitivity to the Magnetic Field

Thus, sensitivity to the magnetic field is important for description of the elastomer behaviour. It is of interest to compare susceptibilities (or permeability) of different types of magnetic elastomers obtained by different methods. The differences between magnetic properties of the elastomers with magnetic particles and magnetic properties of the powder of the same particles can indicate the phase composition of the composite material under investigation.

Sensitivity of the materials are governed by many distinct factors such as permeability of different type of elastomers, percentage of iron particles and voltage supplies. For example, permeability of different type of materials are affected by the geometry of the fabrication and thickness of the elastomers. The volume of iron particles, particle size, particle shape and the alignment of the iron particles inside the elastomer affects the sensitivity of the materials to the magnetic field, which in turn affects the magnetic actuation and the sensing mechanism. In addition, reports experimentally demonstrated that there is a steep difference between pulsating magnetic fields and time-independent magnetic fields and there effects therein [18, 78]. In AC magnetic fields, particles are continually conductive and this will result in the heating up the devices dramatically [18, 78]. The MRE will be heated up by absorbing the AC magnetic fields which results in hyperthermia processes [81, 82]. However, the high intensity of DC magnetic fields lead to the change anisotropic properties and overall matrix elasticity of the hard magnetic fillers inside the elastomer.

2.5 Electrical Conductivity

The current MRE materials are made of Eco-flex silicone rubber or Polydimethylsiloxane (PDMS) which do not conduct electricity [62]. This is due to the fact that the silicone and the oxygen atoms will form strong covalent bond in the giant covalent structure. Thus, these elastomers could behave as good insulators. In our work, the MRE is made of iron particles and silicone rubber. The researcher attempts to make use of the presence of iron particles, which could make the elastomer to be electrically conductive. The researcher makes a perception about the varying the percentage of iron particles will affect the electrical conductivity of the materials. Other than that, the researcher also precepts that the presence of both electrical field and magnetic field will create a motion. This could be further investigated in making a force sensor. The following fabrications have been done to improve the properties of the new materials, which includes the flexibility, sensitivity and electrical conductivity.

2.6 Viscoelastic Property

Viscoelasticity is the property of material that exhibit both viscous and elastic characteristics when undergoing deformations. Viscous materials like honey resist shear force and strain linearly when stress is applied [27]. On the other hand, elastic materials come to their original configurations when stress is been removed. Viscoelastic materials have elements of both of these properties and as such exhibit time-dependent strain where elasticity is usually the result of bond stretching crystal-like graphic planes in an ordered solids [27]. Viscosity, on the other hand, is the result of the diffusion of the atoms and molecules inside amorphous materials. When MREs are exposed to an external EM field, the shear modulus of the material can be represented in a combination of two different parts (i) Shear modulus for its own polymeric nature (G0) (ii) Shear modulus caused due to the external magnetic field (G1). While G0 corresponds the material intrinsic property, G1 changes with the external magnetic field strength [13]. As the external magnetic field intensity increases, the value of G1 increases. This increasing trend signifies higher rigidity and less ductility.

3 Electromagnetically Responsive Soft-Flexible Sensor for Biomedical Applications

Although safe and favourable technology, due to the non-linear relationship between electromagnetic torque and bending angle of the soft material, quantization of the magnetic field inside a deformable structure is still a nontrivial problem to investigate [1]. In this realm, we propose a novel soft-squishy, flexible force sensor approach for active tactile sensation that utilizes soft morphological computation [51, 64]. This research is motivated by hominoid finger's extraordinary combination of fibroblast bone tissue and flexible muscle for grabbing and sensing effective force feedback while gripping a delicate, fragile object in real-time environment [86, 87]. We intend to create an electromagnetically driven tactile sensing system that will be an integration of actuation (magnetorheological paradigm and electromagnetic) and sensing elements (electrical conductivity). The main idea of this proposal will be to have a comparative study with electrical conductivity to address the value of stress generated by the human finger with close proximity. This device when actuated will change its morphology/stiffness and generate electrical stimulus transitions for different posture of embedded sensing. As a result, the proposed device can be proactive in sensing tasks depending upon the EM field variations. Conclusively, this work will be an example of soft morphological control in sensing, and projected to open a new trend in development of tactile sensing system for medical rehabilitation device and therein [90].

The proposed prototype is a three-dimensional structure composed of a sensing coil which is fabricated using natural rubber to be used as a soft-flexible bend sensor for medical applications. The sensing coil mainly acts as an antenna, which is subjected to varying magnetic field and in turn generates electrical signals. The strength of the signal depends on the area of the coil, thickness, number of turns and the magnetic core running through the windings of the coil. The strength of the magnetic field applied to the coil would also determine the strength of the field generated.

FIGURE: Overall Objective of the whole project.

3.1 Material Fabrication

Based on the types of curing process, MR elastomers can be attributed in two distinct categories as (i) Isotropic MR elastomers and (ii) Anisotropic/Aligned MR Elastomers. For each of these two categories, silicon oil, silicone elastomer and carbonyl iron particles are mixed homogeneously to form a viscous liquid. The entrapped air bubbles need to be desiccated minutely in either vacuum chamber (preferably) [37, 83] heat treatment [48]. For isotropic elastomer preparation, the predefined viscoelastic liquid is cured without a magnetic field. On the other hand, anisotropic elastomer needs a higher magnetic field [normally higher 0.8 T] [21, 48, 75] as to align the magnetic particles in some ordered manner.

In our proposed methodology for biomedical force sensor, the copper coils are fabricated in different configurations by varying the number of turns and magnetic permeable core to compare the strength of electric field induced and the inductance of the coil [25, 26, 45]. Three different configurations were tested on the sensing coil having turns of 50, 100 and 150 by changing the core from air to hard grade carbonyl iron micro particles and 3M flexible magnetic strip. The fabrication process involves coiling of 0.25 mm copper wire in the form of solenoid keeping the length, thickness and area of cross section of the material constant for all three configuration. The outer layer of the sensing element is soft and fabricated using a cylindrical mould. The soft material (Eco-flex 00–309a natural rubber) mixture is poured in to the mould. To remove air bubbles, the mould is placed in a vacuum chamber for 5–10min. Next, the mould is placed in an oven for about 90 min at 80 °C for curing. Then, the part is removed from the oven, and the mould is removed. It can also be fabricated by leaving the mould at room temperature for about 4 h. The elastomeric structure formed after curing improves the flexibility of the coil provides a better insulation to copper windings and preserves the shape of the solenoid coil when bent or flexed at different angle. A multiple layer of silicon coating is provided by the same method of fabrication as described above for making the device robust under different working conditions. The working principle of this sensor is based on the displacement of the iron core when bent through a certain angle as shown in Fig. 3. This is because the inner curvature of the coil windings remain in contact and the outer curvature is displaced with respect to the amount of bending.

The coil will have a variable inductance when subjected to bending which can be mapped to the displacement and the angle of bending. By calculating the inductance change based on the displacement the amount of bending can be determined. Hence, the sensing determines the angle of bending by measuring inductance and displacement of the pitch in the helical coil by the application of variable magnetic field [11]. Figure 4 shows the bending in applied magnetic field.

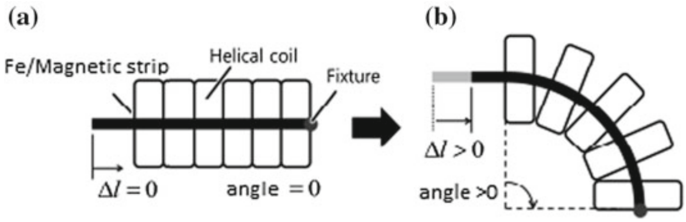

Fig. 3 Sensor working principle **a** when straight **b** when bent.

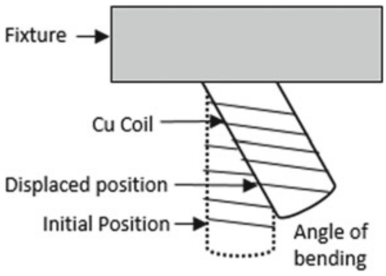

Fig. 4 Bending of the sensing coil in applied magnetic field.

3.2 Summary of Actuations in Soft Robotics and MRE

In the field of actuation in soft robotics, there are different regime of actuations. The pneumatic actuation is still widely used in soft robotics. On the other hand, there are very few well-characterized soft materials, which can be used extensively for the 3D printing—therefore, the field of soft robotics is a fertile land for future chemists to discover new polymers. To make this polymers work for a suitable desired task in most optimum manner, a well-qualitative direct actuation performance comparison is a must choice. In Table 2, we unveil the classification of actuation methods with traditional approaches like motors and cables for the sake of completeness [5].

In our proposed research for MRE-based force sensor, fabrication in making a humanoid hand-like gripper will show the advantages and disadvantages. Based on the prototype, we need to come up with the idea to make smart materials which could show magnetic actuation. In this work, silicone elastomer materials is developed to achieve the bending movement under a magnetic field. These smart materials are different form Magnetorheological elastomer (MRE) or magneto- responsive/active elastomers (MRE) materials. This new type of materials is able to conduct electricity upon actuation. It supposes to change its conductance when a force is applied on the materials. Thus, it is able to work as a sensor in future premises.

3.3 Results and Discussion: An Initial Study

An initial study related to bending flexible force MRE sensor is conducted using the finally fabricated elastomer that consists of a copper coil, iron particles and silicone rubber (details mentioned in the experimental section). In this part, we have tested the voltage difference/drop across the flexible MRE coils when bending the elastomer in different angles. To investigate more into the mechanical property paradigm, we underwent compression and tensile expansion test (Instron UTM), and recorded voltage drop across the elastomer. The experimental setup shown in the

Table 2 Type of actuations and their advantages *and* Challenges (H—High M—Medium L—low) (Data depicted from [5]).

Type of actuations	Motor	Shape-memory alloy (SMA)	Shape-memory polymer (SMP)	Dielectric elastomeric actuators (DEA)	McKibben pneumatic artificial muscles (PAMs)	Fluidic elastomeric actuator (FEAs)
Materials of use	Metals/alloys	Copper–aluminium–nickel and nickel–titanium(NiTi)	Ploy(urethane)-based thermoplastic	Silicone and Acrylic Elastomers	Fibre braid	Synthetic elastomeric films
Operating physics	Cables pulling for flexible bending	Electrical current induced Joule heating	Polymer	Electrostatic force of application	Gas chambers	Chamber network
Stress	M/L	Wires: high; Springs: medium/low	M	M	M	H
Strain	High (H)	M	M	H	H	H
Power density	Medium (M)	H	M/L	M/L	M/H	H
Scaling dimensions	Low (L)	H	H	M/L	M/L, mainly because of the pneumatic pumps	M/L
Response velocity	H	L	L	H	H	M/H
Advantages	Ease in assembly/disassembly, light moving components and low cost	High mass specific force	Less weight per unit volume of material, reversible change of elastic modulus, ease of processibility, lower cost	High strain/stress and mass specific power	Soft fabrication, quick actuation, easily integrated into three-dimensional soft actuated materials	Operated both pneumatically or hydraulically, versatile fabrication approaches to embed fibres
Challenges	Non-linear behaviour, unidirectional constraint that can only pull and not push, undesirable disturbances lead whole system uncontrollable	Relatively low (nearly 5%) strain, force generation in SMAs depends on temperature change, overheating or overstraining can cause permanent damage to the actuator	Low modulus	Requires a rigid frame for prestrains the elastomer, reliability of the compliant electrodes needs improvement. DEAs actuation requires high voltage	To achieve relatively high forces and displacements, they required high power and complex compressed air supply systems, friction between bladder and the mesh also contributes to actuator hysteresis.	Slow actions

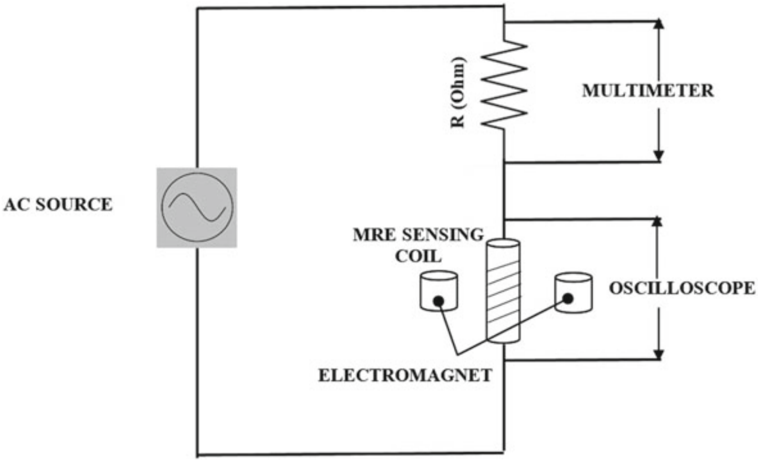

Fig. 5 Schematic of the experimental Setup.

figure includes AC Signal generator, oscilloscope and resistor to prevent our sample from short-circuiting to the supply (Fig. 5).

The principle idea behind this result and discussion subsection of this chapter is to have a comparative study with electrical conductivity to address the value of stress generated by the human finger with close proximity. To illustrate more in this paradigm, in future we will have different materials like Eco-flex, PDMS and conductive foam, Hydrogel etc. to characterize and get the comparative study with the amount of precision it can attain for force sensing. The output of electrical conductivity will be compared with the stress strain analysis (INSTRON UTM) and thus we will reverse engineer the whole system to be rugged and optimal for use in medical rehabilitation further (Fig. 6).

According to the preliminary experimental setup, when MRE elastomer is bent, the average maximum voltage across the elastomer will also change accordingly. The initial length of the MRE elastomer material is 4 cm long and the number of turns of copper coil inside the material varied from 50 turns to 150 turns. For this experiment, the frequency used is about 10 MHz. The following data are obtained from the bending angle and voltage drop across the MRE sensing coil accordingly.

Parameters	Dimensions
MRE sensing coil Length	4 cm
Diameter of the cylindrical MRE elastomer	0.56 cm
Series Resistor R	68 Ω
Supply Frequency	10 MHz

In the following experiments, we have tested preliminarily the change in voltage when bending the elastomer in different angles. In this experiment, the initial position

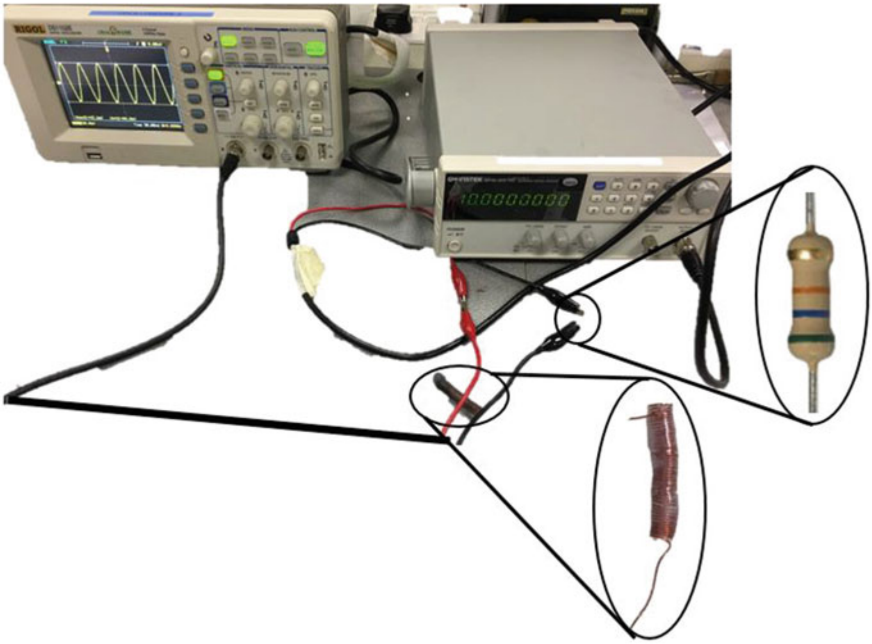

Fig. 6 The experimental setup.

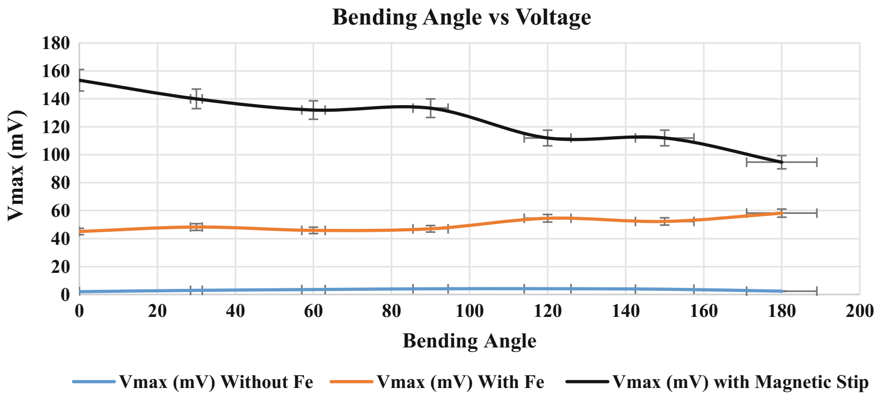

Fig. 7 Bending angle versus voltage—Average value comparison of samples with N = 50.

of the material is in a straight line, at this position, we assume the bending angle is 0 degree. The material will be bent clockwise each time and the voltage across the material will be recorded. Here we have investigated magnetic field strength in correspondence to induced voltage in the coils with three different perspectives as (i) without iron particle (ii) with iron particle and (iii) with introduction of a magnetic strip (Figs. 7 and 8).

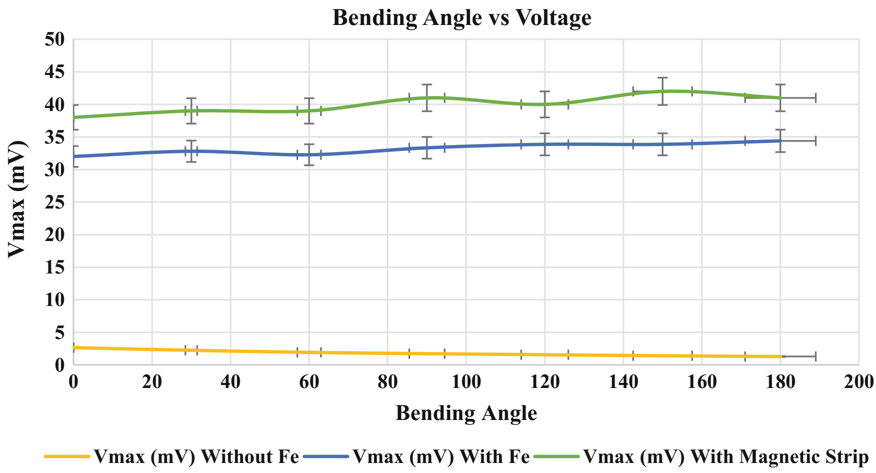

Fig. 8 Bending angle versus voltage—Average value comparison of samples with N = 100.

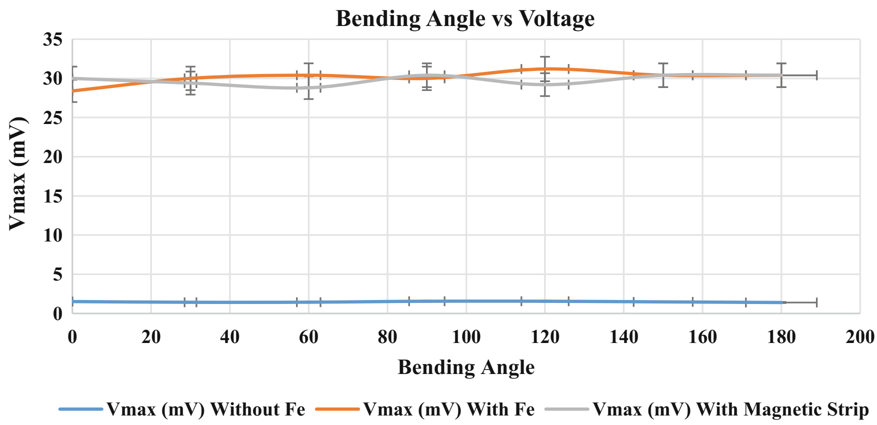

Fig. 9 Bending angle versus voltage—Average value comparison of samples with N = 150.

With the changes in the mechanical property, the EM induction also changes and in turn change the inductance of the MRE elastomers. We have calculated inductance when the coil possess only turn 50 as hereunder (Figs. 9 and 10).

From the above plots, we can extrapolate an understanding that when there is a tensile expansion force applied on the MRE-based sensing coil elastomer, the voltage across the elastomer increases exponentially and it will show a larger voltage difference across the elastomer. However, when there is a tangential compressive force applied on the material, it will cause the decrease in the voltage drop across the elastomer and follow nearly a linear trend. By comparing the plot, it is understood that, when the compressive force is applied to the elastomer, there is a linear drop

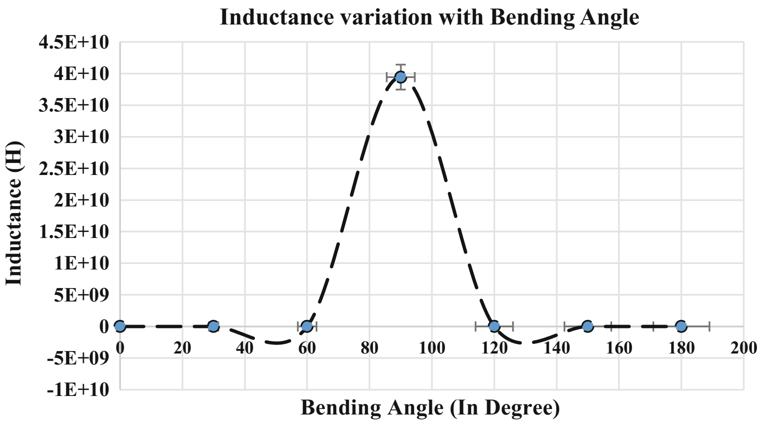

Fig. 10 Bending angle versus inductance—Average value comparison of samples with N = 50.

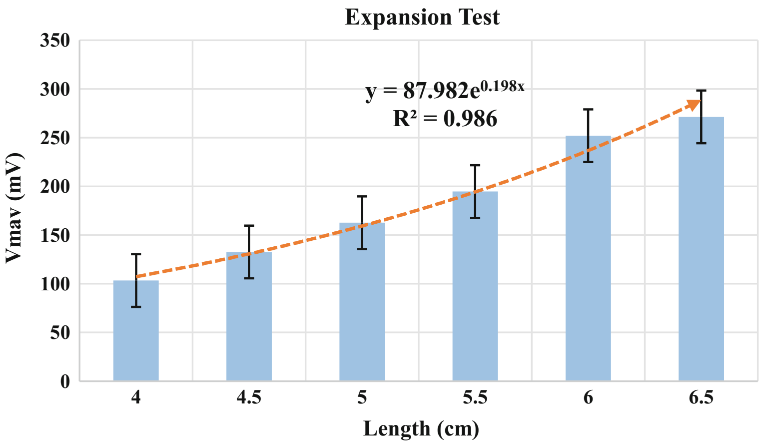

Fig. 11 When there is a tensile expansion force applied on the MRE-based sensing coil elastomer, the voltage across the elastomer increases exponentially and it will show a larger voltage difference across the elastomer.

in voltage as compared to the tension force. This shows that for our fabrication, it is more sensitive to apply compression for it work as a force sensor (Figs. 11 and 12).

3.4 Recommendations and Future Works

In this proposed research, we aimed to fabricate a new kind of MRE elastomers, which could both generate small movements under magnetic and electric field. This elastomer is believed to work as a sensor to detect the bending angles and movements.

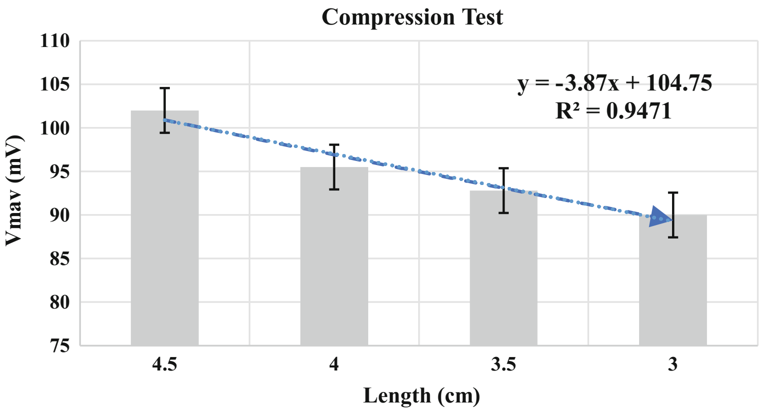

Fig. 12 When the compressive force is applied to the elastomer, there is a linear drop in voltage as compared to the tension force. This shows that for our fabrication, it is more sensitive to apply compression for it work as a force sensor.

The novel approach we have implemented herein is to use copper coil and iron particles for material fabrication to be implemented in a thin cylindrical silicone rubber layer. Compared to normal MRE materials, our approach is sensitive towards both electrical and magnetic stimuli and conductive therein. The fabrication process deployed herein is simpler solution and can be downscaled in the laboratory as per demand. We believe our setup to be conductive for flexible force sensor and thus can be implemented in medical industry and flexible wearable electronics research further. In addition to our preliminary approach the process of fabrication is cost effective and yet resilient to external electromagnetic force. By implementing our fabricated MRE elastomers in rehabilitation glove, it can work as an improvised sensor to help patients to exercise their fingers (Figs. 13 and 14).

In order to increase sensitivity of the elastomer, in future we intend to reduce diameter and hence weight in general for tracking minute differences in results. This in turn will provide a large magnetic field within the coil, which will results in the better bending motions. In addition, the copper wire that is used to make the coil is very soft which could easily deform under as strong force. Thus, a better material with the similar properties of copper wire will be chosen to make the coil. In parallel, we intend to design the MRE sensor embedded with soft skin layer and strain gauges that covers onto another soft substrate sandwiched in between [95]. The results with the strain gauges will characterize the sensor for customized rehabilitation applications. Along with this we are in a process to include conductive ink channel for same design and calculate the change of the conductivity with changing materials from composite to MRE different sample(s).

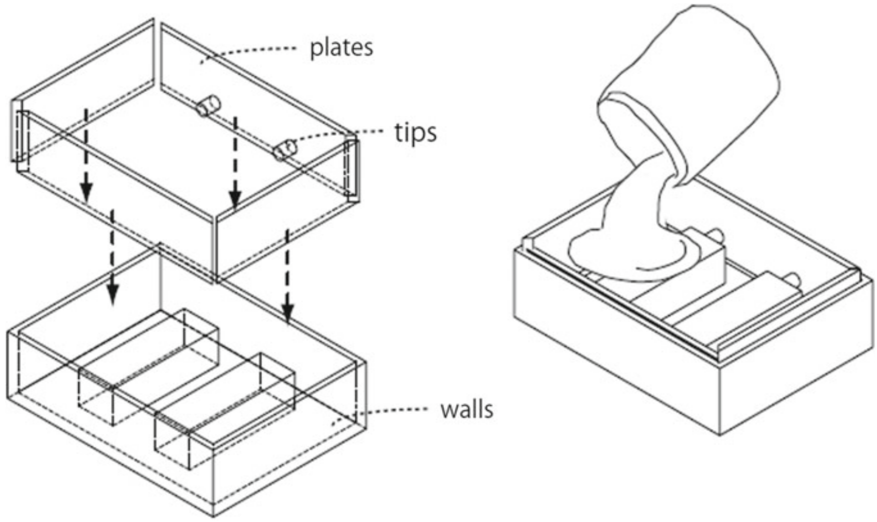

Fig. 13 Design of the sensing system prototype.

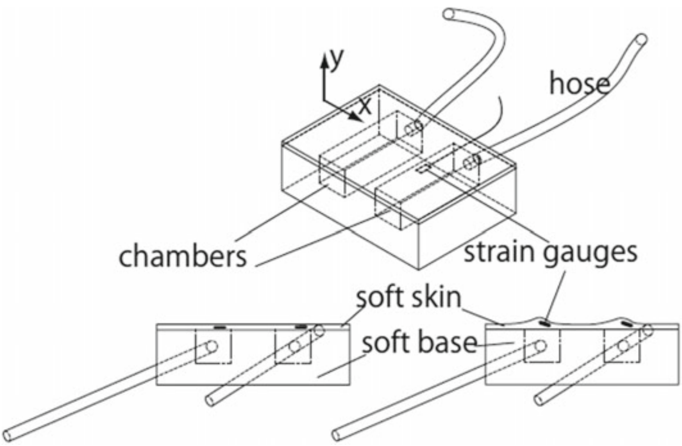

Fig. 14 Design of the sensing system prototype.

4 Proposed Design and Future Perspectives

MRE typically made with iron micro particles ($\sim$5 μm) immersed in silicon can be made softer with silicon oil. Application of a magnetic field can modulate the stiffness of the material [6]. Here exploiting geometry, we propose a saddle shape for MRE. We intend to build the saddle volume around 260 mm^3, which will fit in a 10 mm voxel comfortably. A similar volume cube will be of 6.3 mm voxel, which will exploit higher efficiency in iron particles exposed to the magnetic fields. For optimal control, changing V/V % of iron powder will affect the stiffness properties which further will allow the sample to more efficiently utilize the source B field (Fig. 15).

When we propose to apply magnetic field (B) in one direction, the magnetic micro particles will be attracted and will not only attempt to align with the magnetic field but also be attracted (gradient) towards the source permanent magnet, hence non-uniform fields. This anisotropic attraction will cause tension build up in the saddle and if we were to consider the force measurements in the orthogonal direction, this stiffness of the material should be a result of the tension induced (Fig. 16).

The uniqueness of the design is hypothesized as follows: If we apply a magnetization field in X direction where the edges of the saddle will experience a stronger magnetic field as it moves towards it [7]. Compression and tension in the Z direction should be stiffened due to the tension. The response in Y should be more special, however, compression is resisted while tension is promoted.

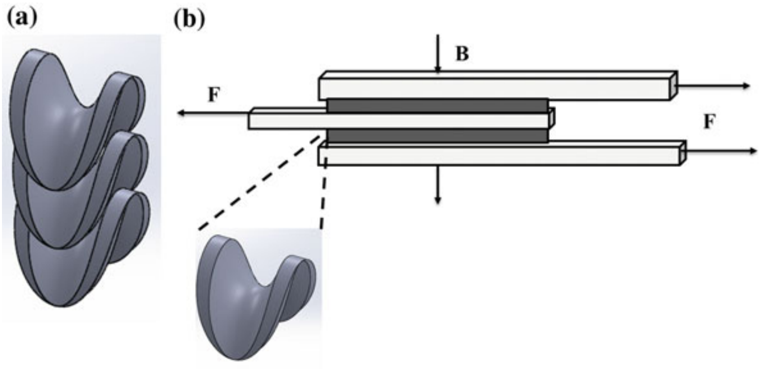

Fig. 15 MRE design exploiting geometry. **a** The design paradigm of the saddle shape MRE. **b** The aligned magnetic particles will attempt to align with the applied fields and resist shearing. Field/Flux alignment is used here to achieve stiffness modulation. The saddle should be able to achieve the same.

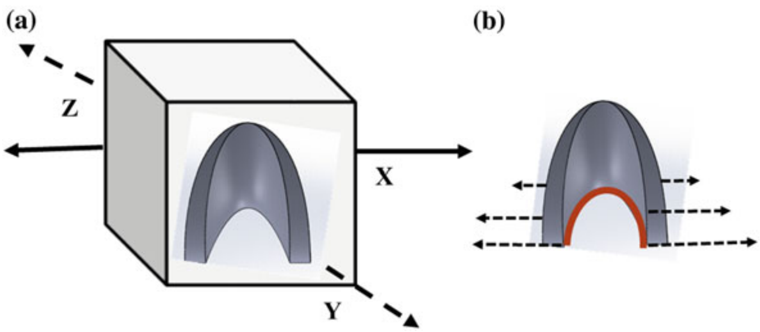

Fig. 16 The force (F) acting on a saddle in a (**a**). Three-dimensional component (**b**). The resultant stress developed in the saddle MRE material.

4.1 Magnetically Modulated/Controlled Soft and Flexible Surgical Robot

So far, the most common mode to control flexible surgical robots is teleoperation, which establishes the mapping between robot motion and movement of a master device [67]. However, the surgical robots can merely be roughly guided to the desired surgical sites with weak obstacle avoidance capability via teleoperation [66]. More precise pose control as well as the motion compensation ability are in demand to strengthen the safety when manipulating the robots during surgical procedures. To better guide the flexible surgical robots to the desired surgical sites and accomplish allotted tasks, the motion planning of the robots can be implemented first to control the robot to roughly reach the surgical sites following an obstacle free path while the visual servoing control mode can be switched on to achieve the finer pose adjustment as well as motion compensation [77]. In addition, the design and motion planning of flexible surgical robots can be assisted by the construction of statistical atlas, which contains information about the average shape and shape variation of a group of patients. This would, further guarantee and improve the performance of robots.

4.2 Magnetic Motion Planning Control (Multiple Targets)

As introduced above, flexible surgical robots are mostly controlled by teleoperation to move from one place to another. However, teleoperation can only control the position of the end effector without considering the movement of the robot body. The unexpected motion of the robot body is likely to cause harm to the patient if the robot body collides with tumours, critical blood vessels, tissues, etc. [59]. Therefore, motion planning is necessary to determine the safest and optimal paths for flexible surgical robots. These robots are desired to manoeuver along the central axes of the cavities in order to maximize the path clearance and better ensure the safety of the navigation. This goal can be achieved by incorporating the centerline extracted from statistical atlas into the motion planning algorithm. Moreover, the motion planning algorithm can be applied using different parameters of the robot [76]. By evaluating the reachability and path quality, the appropriate parameters of a flexible surgical robot can be decided. That is, the motion planning algorithms can assist the optimal designs of flexible surgical robots as well. Furthermore, it is quite hard for some classes of flexible robots, such as the concentric tube robots, to follow the leader. However, by considering the shape constraints of these robots when implementing motion planning, approximate follow-the-leader behaviour of the robots can be achieved. This is quite favourable in minimally invasive surgeries and emphasizes the significance of motion planning. Although only offline motion planning algorithm has been developed in this thesis, the online re-planning process also needs to be worked out in the future based on the real-time feedback from either the visual sensors or electromagnetic (EM) trackers.

5 Forthcoming Challenges and Future Work

A steady hand of a doctor is controlling the machine, with magnets held just above the person on the operating table. When the snake robot is at the right spot, the doctor uses the magnets to release vital drugs right where they are needed.

We are not totally there yet, but the above is not all science fiction—researchers around the world have been developing such soft robots for the past several years. In March 2017, engineers have made a big step forward in controlling soft robots—using magnetic fields to remotely move iron chains made up of tiny particles embedded in the machines. A team of researchers from North Carolina State University (NCSU) describing their research, says that using self-assembling chains that consist of iron micro particles means they can get the simple robots to perform more complex functions [73]. This could one day lead to soft robots used as remotely triggered pumps for drug delivery or robot structures that can be remotely deployed when needed for things like disaster aid, as described by them.

In the near future, we believe the onward research in this soft-flexible robotics regime will be able to advance surgery and medical treatments, while other kinds of autonomous soft robot designs can explore the real world, safely, close to humans, in a way that traditional metal robots cannot. However, researchers still need to solve a set of big challenges: how best to power and control these squidgy machines and many more as described hereunder.

5.1 Material, Design and Fabrication Challenges

The transition from a hard conventional robot to its soft counterpart depends on its underlying materials [85]. The materials, and spatial orientation of the materials, used for soft robots allow for deformable, dexterous, soft interfaces, but the fabrication process poses a challenge for robotic engineers. Currently, soft robotic researchers frequently use state-of-the-art 3D printing and soft lithography fabrication techniques [9, 15, 50, 92], stretchable electronics with wavy circuits [44, 71], soft microfluidic channel with conductive liquid [14], highly stretchable smart textiles, wearable computing, etc.

The next challenge is to 3D print active, multi-material components into a single packet for final use [19, 55, 91]. Though this 3D print technology revolution puts a milestone in the scientific community, it should be mentioned that unlike conventional rigid robots, soft robots require design and manufacturing from scratch, which makes knowledge transfer critical. For example, there are very few well-characterized soft materials which can be used extensively for the 3D printing—therefore, the field of soft robotics is a fertile land for future chemists to discover new polymers. To make this polymers work for a suitable desired task in most optimum manner, a well-qualitative direct actuation performance comparison is a must choice.

5.2 *Computation and Control Challenges*

In comparison with rigid, inflexible, conventional robots, soft robots theoretically possess infinite DOFs making control extremely challenging. The materials used for soft robotics are generally non-linear in nature which makes it exceedingly difficult to predict the empirical coefficient which can mimic the experimental non-linear elastic behaviour, damping coefficients, interfaces between materials, and friction [49]. As soft materials are continuous and deformable [34], the optimal control of the new generation of soft robots needs the state variables of body posture, which are missing since the design parameters are continually changing with deformations, and difficult to address with classical mechanics. The computations become even more difficult if the actuator is to generate optimum force and torque for a desired application. For example, electroactive polymers (EAPs) require very high voltage (in kV range) to operate, while low voltage ionic polymer metal composites (IPMCs) are insufficient to generate enough force and thrust. Pneumatic actuators need an extensive additional pressure infrastructure, while shape-memory alloy (SMAs) has serious trouble with overheating and surface damage. There is ongoing research in simulating continually deformable, highly compliant, flexible bodies using piecewise constant curvature (PCC) model [88], Bernoulli–Euler beam mechanics [31] for deformation prediction or inverse kinematic algorithm. Each of these standard approaches to modelling bio-inspired robotic systems comes with its own challenges. For example, PCC does not necessarily guarantee to incorporate all characteristics of soft robots, for which non-constant curvature model is being introduced recently [68]. On the other hand, inverse kinematics does not include the whole soft body and even the end effectors poses are not included in solutions. Though dynamic modelling somewhat mimics the high deformation of soft materials, interfacing control will be a great challenge as it requires a model-based prediction. So as dynamic modelling progresses, there is a great need to update control algorithms to fit real-time, complex situations.

5.2.1 Control Solutions

Controlling medical devices using magnetic fields is not a new concept and has been explored by a number of researchers. For example, teams at Vanderbilt University in the USA, the University of Leeds in the UK and in Scuola Superiore Sant'Anna in Pisa, Italy, all looked at steering and moving capsule endoscopes [17, 41, 57, 74, 80]. Researchers at the Chinese University of Hong Kong have also magnetically steered micro-robotic swarms [93, 96].

Typically, an autonomous robot must have a basic body structure, sensors, a central control system (microprocessor), actuators (motors), a power supply and a programme for its behaviour. Building a body from soft materials, like polymer, can be done by casting, injection moulding and 3D printing. Sensors and microprocessors can now be manufactured small enough to be embedded inside a soft robot without compromising its flexibility. Electronic components too can be made flexible or even stretchable. However, traditional electric motors cannot be shrunk down and embedded in the same way that sensors can, and they become less powerful the more you shrink them.

Hydraulic and pneumatic systems have been used to control soft robots but they have to be tethered to the machine—not so useful if you want a robot to travel long distances. Other options have been to use so-called soft actuators such as electroactive polymers (EAPs), macroporous gels and other phase-transition materials, but much more research needs to be done before they are effective motors. Some of the most widely used 'soft' actuators use threads of shape-memory alloy wires or foils. These are alloys that change their shape when they are heated, acting similar to muscles. So far, they have been found to be inconsistent, energy inefficient and easily affected by environmental conditions.

5.3 Energy Demand and Optimization Challenges

However, making sure these soft robots have enough power to move outdoors for hours upon end is not an easy task. Anything that is electrically powered must store energy in batteries or capacitors. Although these can be made relatively flexible, they are not widely commercially available and cannot store large amounts of energy. Researchers are instead turning to a biologically inspired solution: storing chemical energy in the soft robot and using it when needed just like fat is transformed into sugars, fats and proteins to provide energy for migrating birds.

The work [73] using magnetic field-driven soft robots is ultimately constrained. It would take incredibly large amounts of magnetic power to move larger soft robots out in the real world.

However, when it comes to autonomous soft robots controlled using the perhaps more promising approach of chemical reactions, the applications could be much wider. Robots with the muscle control and dexterity of animals and humans could be used for handling delicate items in warehouses or creating powerful exoskeletons for the elderly. The technology could even be used to create homes that could morph or change depending on its environment, like a soft robotic wall that could morph into different shapes when needed. Although it may sound far-fetched, with the field of soft robotics making strides every day, it should not be long before we start to see these ideas become a reality.

5.4 Emergence of Biobots: A New Promising Approach

The first development on hybrid bio-robots based on muscular cells have already made appearance in the form of proof-of-concept Muscular Thin Films (MTF) [79] or systems that can crawl, swim or grip [3, 20, 61, 63, 65, 89]. Most of these devices are based on cardiac muscle cells, which have the ability of self-beating, therefore reducing the need of a control mechanism [24, 89]. Nevertheless, a control of the frequency of beating can still be achieved by electric field or genetically modifying the cardiac cells to respond to a light stimulus [10, 63]. Cardiac muscle cells provide a suitable solution for proof-of-concept bio-robots or biomedical studies in heart-like constructs or organ-on-a-chip devices, since they are already differentiated into striated muscle cells. However, they are not the most convenient election for hybrid bio-robots, mainly because they contract continuously without the capacity of halting. Skeletal muscle cells do not present this problem and therefore are thought as a potential solution for hybrid soft robotics, but they have been proven more complicated to use than cardiac cells. They need to be seeded as myoblasts and differentiated into myotubes in a process that lasts several days. More developed and striated muscle tissues require the formation of sarcomeres for an optimal force generation.

Skeletal muscle cells have been used in MTF and in simple crawlers or swimmers [20, 65, 79]. However, none of these actuators discuss the formation of sarcomeric structures in their constructs, leaving only certain basic research studies to deal with that [29]. It has been shown that surface stiffness, electrical stimulation or mechanical strain can improve the development of sarcomeres [22, 23]. For instance, more mature sarcomeres can be obtained when the elastic modulus of the substrate is closer to that of real tissue. In relation to this, it has been reported that a second layer of myocytes can form well-organized sarcomeres, thanks to the mechanical compliance with the first layer of tissue [32]. Besides this, a 3D study of the formation of myotubes and striated muscle in 3D is still at a very early stage. In this regard, 3D bio-printing becomes a unique tool for the fabrication and study of well-developed muscle tissue in three dimensions as well as its integration with printable materials. Moreover, the latter could lead to the fabrication of 3D bio-robots that can produce greater power than their 2D-seeded counterparts. The state-of-the-art 3D bio-printer from RegenHU available at the institution and CELLINK's Inkredible+ 3D bio-printer already in possession by our group provide us with the necessary means for the fabrication of complex 3D Biobots. Furthermore, we will use this technique to pursue forward the study of 3D alignment of myotubes and sarcomere assembly, taking it from basic research towards real applications in the field of hybrid soft robotics and biomedical applications.

Due to the importance of the mechanical properties of substrates, a great deal of research has been carried out in the development of new hydrogels for the support or encapsulation of different kinds of cells. These materials often offer tunable mechanical properties by changes in their composition or cross-linking mechanism, making it possible to reach a better compliance and tissue-like stiffness [33]. They have also been nano-engineered with the addition of magnetic nanoparticles [94], nanofibers [43] or carbon nanotubes [2] to improve the alignment or excitation of myotubes, but it has not been pursued towards an application. 3D bio-printing has been used to obtain 3D structures of different kinds of cells encapsulated in hydrogels that are later crosslinked and it has been proven to be a biocompatible process, as well as suitable for cell survival, proliferation and differentiation [54]. Furthermore, complex architectures can be created with the help of sacrificial materials, such as hydrogels that can be removed after the construct has been crosslinked [38]. So in nutshell, the progressive research in biological cell energy stimulator will be an immediate alterative to high end energy demand and collectively in future be used as an energy harvester.

5.5 Commercial Challenges

Soft robots have great potential to be applied in industrial automation and health care realms. As per the prediction projected by ABI research, the market of personal robots may undergo a sharp fall from previous estimates to $ 6.5 billion by 2017 [53] which leaves engineers no choice but to focus more on cost-effective, flexible soft robots. However, this emerging technology has many challenges to overcome before it can be widely commercialized. First, it is extremely difficult for start-ups to sell directly to end users or to collaborate with large manufacturers, as they typically want to collaborate with companies, which already demonstrate an operational record of accomplishment and financial stability. Second, even if there is financial support for R & D and marketing activities, obtaining regulatory approval for medical applications takes a long time (5–10 years), which means investors need to commit to financial support for a long period.

There is no doubt that in the near future, market demand will influence the commercialization of soft wearable machines and medical robotics. Apart from manufacturing scalability, consumer interests and efficient and viable technical solutions, soft robots will face great pressure to be inexpensive and optimized to meet the steep market demand. The great advancement of electronics integrated chip (IC) and microfluidics network in the last few decades follow Moore's law nicely, while soft robots have yet to climb that ladder and face great challenges in the future. However, according to Bank of America Merrill Lynch (BAML) research, medical robots business is expected to grow at a much faster rate to reach $ 18b by 2022 even though compared to industrial robots, medical robots adoption is still at an early stage, with 1,224 units sold for $ 1.3b in 2014, only accounting for 5% of total robots sold [56]. On a final note to sum up, Prof. George Whitesides—in an interview with Prof. Barry Trimmer—clearly explained that the field of soft robot actuators is still in its nascent phase and will be commercially challenging for researches to explore in the near future [84].

6 Concluding Remarks

Inspired by the exceptional competence of natural organisms, researchers have been widely exploring the possibility to mimic the rich multifunctionality of soft biological species. The motivation behind this gradual shift towards soft robotics derives from the aspiration to tackle more complex, unpredictable environments, which demands higher order of mechanical intelligence. Imitating the inherent traits of their natural counterparts, these futuristic robots are designed to match the elastic and rheological properties, to facilitate easy movement through awkward surroundings. In general, soft robotics offer better mechanical compliance with biological systems than their rigid counterparts. Their biocompatibility, adaptability and capability for developing multiple and complex functions such as actuation in confined spaces, deformability, adaptation to the environment or manipulation of unmodelled objects, account for their potential towards several different applications, including biomedicine.

It is evident by now that magnetically responsive untethered soft robots bring a new way to look into robotics for future generations, which will attract investors and companies for product commercialization. It is also to be noted here that soft robotics in general for whichever actuation is posed upon, offer the potential not to compete with conventional robotics, but to tackle a set of problems that existing technologies have not been able to solve. The vast biomedical applications of soft robotics in rehabilitation, tissue engineering, soft biological cell biology, flexible surgical manipulators etc. are overwhelming and call for a serious investment in research focused on the fabrication and material synthesis of flexible, dexterous and cost-effective cross-linked polymers. There are examples of soft robotics revolutionizing areas beyond biomedical research like disaster management, rescue operations and field exploration. In all these preamble application avenues, magnetic-driven softbots are highly acknowledged as per their wireless, minute control manoeuvre. The question that stands now is whether innovations in rapid prototyping techniques like soft lithography and 3D printing will allow manufacturers to print an entire robot that will be inexpensive, easy to use and satisfy market demand. We can be optimistic to make this new technology grow faster and bring out a sustained environment in which human–robot interaction can rise to a new level.

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