Sensing and Actuation in Monolithic Electroactive Elastomeric Bodies

by

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"When do you think you can submit your thesis?"

T. Giffney, April 2024

"Today."

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Abstract

College of Engineering
Mechanical Engineering

Doctor of Philosophy

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Some of the world's most advanced technology is rigid due to various factors such as; manufacturability, miniaturisability, physical linearity, and more ideal physics in general. In parallel industries is also looking to use automation to improve and replace laborious tasks whether they be domestic, commercial, or industrially related tasks. Their is a growing need for new innovations in technology to utilise the soft robotic solutions that mimic biological solutions seen in nature. This thesis is part of many to improve an understanding of the electroactive polymer subset of soft robotics and the limitations of specific implementations of artificial skin and artificial muscle technologies.

This thesis explores the integration of Electrical Impedance Tomography (EIT) with advanced soft sensing technologies, focusing on carbon black silicone rubber (CBSR) elastomer composites and Dielectric Elastomer Actuators (DEAs) to enhance pressure mapping, strain sensing, and actuation.

CBSR elastomer composites, noted for their high stretchability and biocompatibility, were investigated to understand their resistance relaxation behavior. This research contributes to optimizing the design of flexible dynamic strain sensors by modeling the response of resistance to transient strain inputs. The study developed an EIT-based pressure mapping system using a silicone CB nanoparticle sensing domain that mimics pressure mapping qualities human skin. This system was evaluated for its spatial and temporal resolution, showing potential for creating artificial pressure-sensitive skin with practical applications. Furthermore, the integration of EIT with DEAs was examined to improve the mapping of compressive forces across electrode surfaces. Despite some trade-offs in accuracy due to electrode compliance, this approach offers promising advancements for applications requiring precise actuation and pressure mapping. This work has majorly contributed towards filing a patent for an DEA-EIT actuator-sensor device. Additionally, the research uncovered unintentional power generation in DEAs, which could function as Dielectric Elastomer Generators (DEGs) due to mechanical strain. This finding highlights the dual functionality of DEAs and suggests opportunities for energy harvesting applications. Finally, a portable, low-cost EIT-based hardware system for pressure mapping was introduced. This system enables comprehensive characterization of various sensing domains and supports advancements in EIT-based soft sensor technology, with implications for biomedical devices, robotics, and energy harvesting.

Overall, this research advances the field of soft sensors by integrating EIT with innovative materials and technologies, providing new insights and applications in dynamic sensing and actuation.

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- Tess - for general oddities, support, and love.
- Friends - for constantly questioning "so when are you finishing again?"
- Family - for being supportive.

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Abbreviations

ADC	**A**nalog-**t**o-**D**igital **C**onverter
CAD	**C**omputer **A**ided **D**esign
CB	**C**arbon **B**lack
CFA	**C**artesian **F**orce **A**pplicator
CE	**C**ompliant **E**electrode
CoM	**C**enter of **M**ass
DE	**D**ielectric **E**lastomer
DEA	**D**ielectric **E**lastomer **A**ctuator
DEG	**D**ielectric **E**lastomer **G**enerator
DUT	**D**omain **U**nder **T**est
EIT	**E**lectrical **I**mpedance **T**omography
ERT	**E**lectrical **R**esistance **T**omography
FEA	**F**inite **E**lement **A**nalysis
FEM	**F**inite **E**lement **M**odelling
FPC	**F**lexible **P**rinted **C**ircuit
IDF	**I**oT **D**evelopment **F**ramework
MUX	**M**ultiplexer
PCB	**P**rinted **C**ircuit **B**oard
PCBA	**P**rinted **C**ircuit **B**oard **A**ssembly
PDMS	**P**oly**d**i**m**ethyl**s**iloxane (AKA silicone)
PNEC	**P**iezoresistive **N**anoparticle **E**lastomer **C**omposite
SMU	**S**ource **M**easure **U**nit
SMD	**S**urface-**M**ount **D**evice
SR	**S**ilicone **R**ubber
THT	**T**hrough-**H**ole **T**echnology

Symbols

A	Area	[m^2]
C	Capacitance	[F]
ε	Permittivity	[Dimensionless]
K	Bulk Modulus	[Pa]
ν	Poisson's Ratio	[Dimensionless]
Q	Electrical Charge	[C]
U_E	Electrical Potential Energy	[J]
U_ϵ	Elastic Potential Energy	[J]
R	Resistance	[Ω]
σ	Stress	[Pa]
S	Strain	[Dimensionless]
V	Voltage	[V]
Y	Young's Modulus	[Pa]
z	Thickness	[m]

Dedicated to tinned baked beans in all their glory...

Chapter 1

Introduction and Motivation

Rigid robotic systems often have multiple rotary motors and various sensors integrated together for precise control of the robot, this is mirrored in biology with the animals having many actuator units in the form of muscles and a multitude of various receptors for sensing their environment. The rigidity of rotational motors is stifling creativity in the creation and development of devices amongst many other unforeseen future technology. Engineers are often constrained to solving problems and designing solutions using typical rigid sensors and actuators due to their current ubiquity and their evolved increased efficiency. With the rise of research into soft sensor and actuator devices, these such device need to follow suit of the traditional rigid sensors and actuators and become ubiquitous and viable option for general and specialised engineering design solutions.

This thesis has developed methods and tools for creating and characterising artificial pressure sensitive skin technology. The thesis then continues to explore the integration of this artificial skin technology into an artificial muscle technology. The work in this thesis has ultimately contributed towards a patent for DEA-EIT actuator-sensor technology in a quest to bring this work out of the academic realms into real-world applications.

1.1 Why Go Soft and Not Rigid?

The requirement for soft robotics in general has been driven by the limitations of current rigid robotic solutions to interact with natural organic material. Manipulation of natural organic objects such as animals, plants, fruit, vegetables, and meat have traditionally been handled by humans by hand due to our ability to use our dexterity and intelligent control systems to ensure minimal undesirable damage. With the advance in in technology in various soft robotic actuators[**? ? ? ? ?**], sensors[**?**], and soft robotics control[**? ?**]. The use of soft robotics in place of rigid alternatives, amongst other benefits, has the opportunity to be more sustainable by decreasing waste products during fabrication, using biodegrabale or recyclable materials, shelf life, and use of renewable resources[**?**]. The use of soft robotics brings opportunity of creating devices with a reduced bill of materials size and less moving parts for maintenance. The use of soft robotics in biomedical and aerospace applications is especially desirable due to the difficulties experienced when designing with regular motors in the outer space and near sensitive biological tissue environments such as heat dissipation, lubrication, and mass[**? ? ? ?**].

The most common rigid actuator is the rotary electric motor and the global market was valued at USD 142.2 billion in 2020, with a predicted growth rate of 9.5% until 2032[?]. Although this market is dominated by automobiles which currently require the traditional form of rotary electric motors, growing sectors of this market such as medical, factory automation, and aerospace have potential interest in adopting soft actuator alternatives for the reasons given above. In parallel, rigid strain sensors of types metallic foil and semiconductor, was given a global market value of USD 190.66 million in 2022 with a compound annual growth rate of 3.9% until 2029[?]. Adjacently the pressure mapping global market value, focused mainly on the health sector, was valued at USD 480 million in 2023 with an expected growth rate of 5.1%[? ?]. Many soft actuator technologies could be used in these growing medical, aerospace, factory automation, and agricultural sectors.

Soft robotic actuation can be achieved through various mechanisms including thermal, electrochemical, fluidic, magnetic, and electrostatic. Similarly soft stress-strain sensing can be achieved through various physical principles such as resistive, capacitive, magnetic, and optical sensing methods. Often the function of soft actuators can be inverted such that the deformation of the actuator can produce a signal used for self sensing, in electroactive polymer (EAP) technologies such as dielectric elastomer actuators (DEAs)[? ? ? ?] and ionic polymer-metal composites (IPMCs)[?]. EAPs have the benefit of electronic control over other soft actuator and sensor technologies controlled by fluids, heat, or light which contain the complexity of another energy transfer process.

Proprioception in artificial muscle technology has been made a reality. This is seen in the self-sensing of one dimensional strain of DEAs usually through capacitive measurement between the compliant electrodes during operations to obtain the magnitude of a contraction. However, the pressure mapping done similar to the mechanosensation performed by cutaneous mechanoreceptors on an artificial muscle device has not been explored as of writing this thesis.

Publications towards this thesis include three conference papers, one journal paper, and one provisional patent filed. This thesis has converged on the use of conductive particle based elastomer composites and their use in sensors and actuators, in particular an electrical impedance tomography (EIT) based artificial skin and it's integration into the artificial muscle technology, dielectric elastomer actuators. The composite type used throughout the thesis is simple to fabricate but not well understood in terms of its electromechanical transient and dynamic characteristics. The modelling of such conductive particle composites would elucidate the feasibility of inverting the model to create a responsive strain sensor. This composite has been characterised in one-dimension several times in literature already however, if a two dimensional sensing application of this composite is desired the characterisation of the sensor in two dimensions must be completed. A method to do such 2D sensing is using EIT. EIT has been used in the past for a huge range of applications, with few exploring the use of EIT as a pressure mapping sensor. Although EIT-based pressure mapping was first discovered 30 years ago, the technology is still in its infancy with several problems needing to be resolved before the technology can be used reliably in real-world applications.

1.2 Research Objectives

The research objectives and questions for this thesis are given below:

1. Quantify and analyse static, dynamic, and transient phenomena seen in conductive particle composites.

2. From the characterisation in objective 1 mitigate the effects of the transient phenomena.
3. Create a set of metrics for quantifying the performance of an electrical impedance tomography based artificial skin.
4. Simulate and integrate an electrical impedance tomography based artificial skin onto a dielectric elastomer actuator.
5. Investigate the energy harvesting of a device that is both a dielectric elastomer actuator and electrical impedance tomography device.

1.3 Chapter Contributions

Chapters **??** - **??** contain the core novel research contributions. Chapters 2 and **??** provide essential background knowledge and future research directions for the thesis respectively.

Chapter 2 - Literature Review: This chapter explores the nature of biological skin and muscle from an engineering perspective, quantifying necessary functions and properties desired to replicate or supersede for their artificial equivalents. The thesis then describes state-of-the-art soft sensors and actuators and their function.

Chapter 3 - A Simple Conductive Elastomer Composite Material with Complex Behaviour: This chapter uncovers the electromechanical tensile and compressive properties of carbon black silicone composites, in order to understand the material before it's use in sensors and actuators.

Chapter 4 - An Improved an Electrical Impedance Tomography Based Artificial Soft Skin Pressure Sensor: This chapter discusses the use of electrical impedance tomography to create a pressure mapping sensor and provides tools for analysing the suitability to various applications and choosing a suitable sensing domain.

Chapter 5 - Giving Artificial Muscles the Sense of Touch: This chapter describes the integration of the pressure mapping technology discussed in the previous chapter, how it can be integrated into dielectric elastomer actuators, and the trade-offs.

Chapter 6 - Unintentional Power Generation in a DEA-EIT Sensor-Actuator Device: This chapter discussed the unintended power generation of the simultaneous sensor actuator device discussed in the previous chapter.

Chapter 7 - A Portable Electrical Impedance Tomography Based Pressure Mapping Sensor and Validation System: This chapter discusses the small form factor, low-cost hardware design for a hybrid artificial muscle - artificial skin based device.

Chapter 8 - ?? Modelling of DEA-EIT Capacitively driven Hybird Sensing and Actuation Device: The is chapter models the a DE-EIT device in order to find an optimal range of parameters at which capacitive shunting can be used to improve the DE-EIT pressure mapping device responsiveness.

Chapter 9 - The Biomimmetic Re-Evolution: This chapter discusses the future direction of the technology discussed in the thesis and acknowledges the future of the broad field of soft robotics.

Chapter 2

Literature Review

To replace and supersede tasks that can currently only be performed by humans due to their dexterity, physical makeup, and intelligence; the skin and muscles completing these tasks can first be understood and quantified. Subsequently a review of various electrically driven artificial skin and muscle technologies was completed. Finally, background theory on piezoresistive elastomer composites and two specific technologies of soft sensing and actuating devices is given to setup a foundational knowledge base and reference for the rest of the thesis.

2.1 Biological Skin form and function

Skin is the largest organ in the human body with many functions, however this thesis only aims to replicate some pressure-sensitive functions of skin. Two pressure-sensitive categories of skin and muscle tissue transducers which allow for dexterous manipulation of objects are:

1. Proprioceptors: respond to internal mechanical stimuli in a joint capsule, tendon, or muscle to give the sense of motion.
2. Cutaneous mechanoreceptors: respond to mechanical stimuli usually external to the body, including pressure and vibration, for the localisation of sensations.

Locations of both proprioceptors and cutaneous mechanoreceptors are shown diagrammatically in Figure 2.1. Proprioceptors aid in determining pose estimates of body parts in space, acting as sensors providing feedback closed-loop control for the neurological motion control of body parts. Whereas cutaneous mechanoreceptors have various roles including object recognition, manipulation control, as well as motion control.

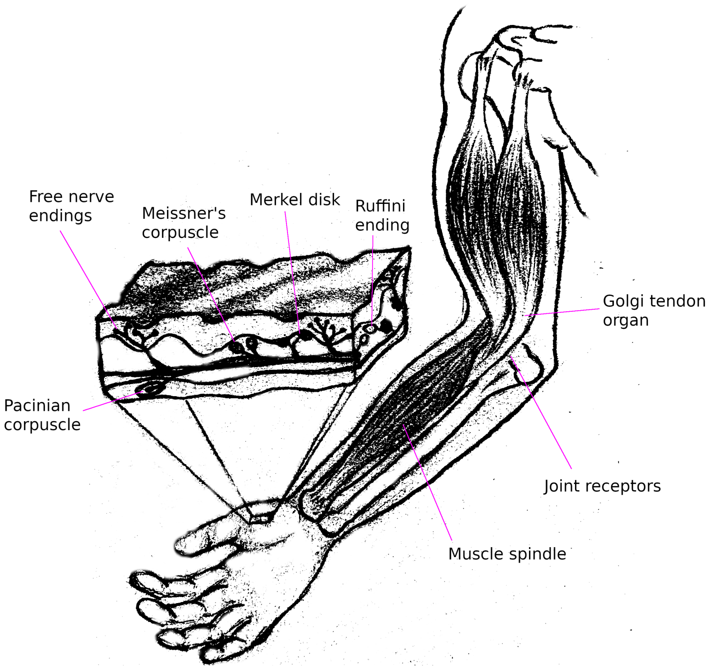

FIGURE 2.1: Examples of the locations of proprioceptors and cutaneous mechanoreceptors in the human body.

The function of both kinds of receptor have been mimicked by certain device technologies. For example proprioceptors have been mimicked in wearables and human assistive devices where joint motion has been estimated by sensors such as rotary/linear encoders, IMUs, and stretch sensors fixed adjacent to joints to calculate pose estimates of limbs[? ? ? ?]. Examples of such devices are displayed in Figure 2.2

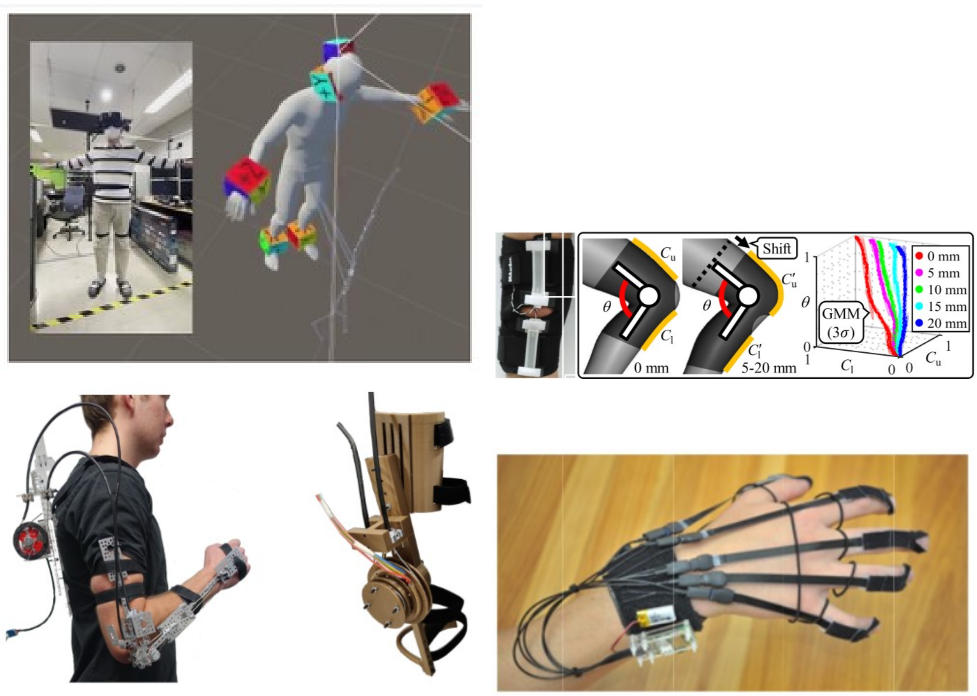

FIGURE 2.2: Clockwise from top left: IMU pose estimation[?] (© 2022 MDPI), stretch sensor knee joint pose estimation[?] (© 2020 IEEE), encoder elbow pose joint estimation[?], stretch sensor hand joint pose estimations[?].

Cutaneous mechanoreceptors have been mimicked by the development of pressure mapping of flexible surfaces. Examples of such technologies include, foot pressure based gait analysis, wheelchair seat pressure mapping. Examples of these sensors are shown in Figure 2.3.

FIGURE 2.3: Various pressure mapping devices. From top-left clockwise: Xsensor wheelchair pressure mapping sheet[**?**], Pressure Profile Systems pressure sensors on a robotic hand[**?**], Soft pressure mapping gripper[**?**], Tekscan thin pressure mapping platform[**?**], Tactilus seat pressure mapping system[**?**]

Many of these pressure mapping technologies don't accurately mimic desirable qualities of regular biological skin and are specialised for their specific use cases. The following sections quantify characteristics of pressure sensitive skin.

2.1.1 Skin Construction and Types

Skin is a laminate structure consisting of three main layers, the epidermis, dermis, and hypodermis. The top two layers the epidermis and dermis are a subset of the cutaneous layer which contain the majority of the pressure-sensitive mechanoreceptors [].

The skin can be categorised as glabrous/hairless or non-glabrous/hairy. Glabrous skin contains many of the mechanoreceptors given in Figure 2.1 whereas non-glabrous skin will also contain C-tactile afferent receptors for obtaining sensations through hair follicles. However this work is exploring simple monolithic bodies so will not be replicating the sensor function of non-glabrous skin.

Depending on the region of skin different force resolution and spatial resolution will incur. The tensile properties of skin is governed by skin tension lines, also called Lager's lines, which show the direction in which the maximal stretch can occur.

Cutaneous mechanoreceptors and their functions are given in Table 2.1.

Table 2.1: Comparison of typical mammalian mechanoreceptors characteristics [?].

Receptor	Meissner corpuscle A1	Ruffini Corpuscle A2	Pancian Corpuscle B1
Perceptual sensory functions	Skin movement, handling objects	Skin stretch, movement direction, hand shape, and finger position	Fine tactile discrimination, form and texture perception
Skin stimulus	Dynamic deformation	Skin stretch	Indentation depth
Localisation	Dermal papillae	Dermis	Basal layer of epidermis / around guard hair
Conduction velocity	35 - 70 m/s	35 - 70 m/s	35 - 70 m/s
Receptive field	22 mm^2	60 mm^2	9 mm^2
Receptor density	150 / cm^2	10 / cm^2	100 / cm^2

2.1.2 Characterising skin

The sensing qualities of skin is crucial for the sensory feedback in complex manipulation tasks. To aid the creations of technology that mimics qualities of biological pressure sensitive skin, the mechanical properties of must be characterised. Biological human skin is highly variable in terms of its mechanical and sensing properties depending on the region of skin, giving large variation in skin characteristics. Skin can be characterised in terms of the following mechanical characteristics:

1. Elastic modulus - The static elastic properties determined by a linear region of stress and strain of the material [Pa]

2. Storage and loss modulus - The dynamic elastic and viscoelastic properties determining the relationship between stress and strain [Pa]

3. Ultimate tensile stress (UTS) - The maximum tensile stress that a material can tolerate before breaking [Pa]

4. Life cycle - The time or number of actuation cycles in which it takes for the actuator to degrade such that it cannot perform its intended purpose to specified standards

5. Viscoelastic creep and relaxation - All viscoelastic materials will experience strain creep and stress relaxation to varying degrees depending on the viscoelastic properties of the material [$mm.s^{-1}$ and s]

6. Skin thicknesses - the thickness of all layers of skin the cutaneous epidermis and dermis and thickness of the hypodermis [mm]

7. Skin surface area - Biological skin has a large surface area and can also be regionalised to map skin function and sensitivity [m^2]
8. Isotropy/Anisotropy - The directionality of skin properties, also known as skin tension lines, give a topological map of the maximal stretch (i.e. minimal elastic modulus) direction of regions of skin.

Some of the functional properties in terms of pressure mapping include:

1. Spatial resolution and touch acuity - The spatial resolution of biological skin, which is mainly dependent on the innervation, mechanoreceptors density, and thickness of the cutaneous layers of skin [**?** **?** **?**]
2. Static force resolution - This is the detection resolution of static or slow-acting forces acting upon the skin [**?**]
3. Temporal resolution - This is the detection resolution of fast-acting forces acting upon the skin often required for texture recognition [**?** **?**]

A quantitative characterisation of mechanical and pressure sensing functional skin properties include:

1. Elastic modulus - varies largely depending on test method, test skin type, and subject. Values found in literature include 83.3 $\pm$ 34.9 MPa [**?**], 0.1 - 2.4 MPa [**?**], and 10.4 - 89.4 kPa [**?**].
2. Storage and loss modulus - varies largely depending on test method, test skin type, and subject. Values found in literature range include 141.9 $\pm$ 34.8 Pa and 473.9 $\pm$ 42.5 Pa at 0.8 Hz [**?**], 473.9 $\pm$ 42.5 Pa and 32.3 $\pm$ 10.0 Pa at 205 Hz [**?**].
3. Ultimate tensile stress - 21.6 $\pm$ 8.4 MPa [**?**]. 28.0 $\pm$ 5.7 MPa [**?**]
4. Life cycle - Skin cells are constantly growing, dying, and shedding. Skin is always actively remodelling based on external stimuli.
5. Strain creep - The strain creep was found to be 2.7 kPa.s for a 10 Pa step input on a dermis skin sample [**?**].
6. Skin thicknesses - The thickness of human cutaneous skin ranges from 0.6 to 2.6 mm with an average skin thickness of 2 mm [**?**].
7. Skin surface area - The average surface area of skin in adult humans is 1.7 $\pm$ 0.1 m^2 [**?**].
8. Isotropy/Anisotropy - The tension lines in skin are determined by collagen fibre orientation and dynamic stretch events [**?** **?**]. The elastic modulus of human skin was reported to be 160.8 $\pm$ 53.2 MPa parallel to the skin tension lines and 70.6 $\pm$ 59.5 MPa perpendicular to the tension lines [**?**]. The UTS of human skin was reported to be 28.0 $\pm$ 5.7 MPa parallel to the tension lines and 15.6 $\pm$ 5.2 MPa perpendicular to the tension lines [**?**].

1. Spatial resolution and touch acuity - The tactile field area increases with indentation depth for certain mechanoreceptors with a range of 5 - 12.6 mm^2 [**?**]. Two point discrimination is another metric for determining spatial resolution an has been determined as 3.7 $\pm$ 0.7 mm [**?**]. The receptive field varies depending on the mechanoreceptors used so has been reported to be between 1 and 60 mm^2 as another methods of inferring spatial resolution [**?**].

2. Force resolution - Minimum force detection on various regions of human skin was found to be between 67 - 1007 mg [?], and various mechanoreceptors 0.73 - 122.6 mN [?].

3. Temporal resolution - Depending on the mechanoreceptor sensing the force input, a frequencies ranges of 0 to 800 Hz can be perceived by human skin [?]

2.1.3 Skin Modelling

Developing robust mechanical models for human skin is non-trivial for three main reasons:

1. high degree of viscoelasticity
2. regenerates and heals
3. made from various types of cells in a laminate structure

To solve the complexity of modelling such a material a review by Landry et al.[?] shows that many researchers have applied various non-linear mechanical models including Ogden, Mooney–Rivlin, Neo-Hookean, Yeoh, Humphrey, and Veronda–Westmann. When recreating an artificial muscle it is desirable to minimise the mechanical material model complexity so that the material can be more easily integrated into a control system with known behaviour. Similar modelling techniques can be used to model conductive particle elastomer composites due to the similar hyper-elastic and visco-elastic behaviours observed.

2.2 Pressure Mapping Artificial Skin Devices

This section will be outlining some of the main technologies which are flexible and/or soft and can map force events throughout a two dimensional surface. A particular focus on electro-active polymer (EAP) based sensing is present due to the potential of miniaturising the technology and the range of miniaturised electronics currently available. Electroactive polymers are essentially polymer materials which can be used as transducers which change electrical properties based on a mechanical input, vice versa.

2.2.1 Soft Pressure mapping technology

Pressure mapping devices can be categorised into their various sensing technology, such as resistive, capacitive, inductive, magnetic, optical, and acoustic. Examples have been gathered by [] showing the limits and trade-offs between each sensing technology.

2.2.1.1 Resistive

Soft resistive pressure mapping has been commonly achieved in the past by using arrays of piezoresistive sensor elements []. The resistive elements can be made using several different flexible piezoresistive materials.

- Conductive particle polymer composites [? ? ?]

- Intrinsically conductive polymers [? ?]
- Microfluidic metals [? ? ?]
- Hydrogel structures [? ? ?]
- Flexible piezoresistive semiconductors [? ?]

	Conductivity	Piezo-resistivity	Change stiffness	Fabrication	Cost	Environmental Stability	Toxicity
Intrinsically conducting polymers	- Dependent on polymer used. (S/cm [?])	- Dependent on polymer used - Exponential relationship with strain[?]					
Electrolytic hydrogels							
Conductive particle polymers					$		
Conductive particle paste					$		
Conductive textiles				Complex			

A commonly used piezoresistive material is conductive particle polymer composites.

2.2.1.2 Capacitive

Similar to resistive pressure mapping, capacitive pressure mapping has more commonly been done using arrays of capacitive elements. Many capacitive touch sensors use the human body to shunt the electric field between the capacitor electrode(s) to a common ground. However the operating principle of capacitive-based strain sensors rely on the deformation of the capacitor dielectric and/or the capacitor electrodes.

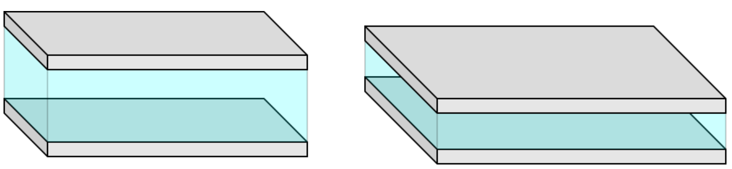

FIGURE 2.4: Two grey electrodes across a blue dielectric medium. Left: Uncompressed state. Right: Compressed state exhibiting larger electrode areas and a thinner dielectric thickness.

2.2.1.3 Magnetic

Magnetic strain mapping devices can be achieved using several methods. One method is to have a three layer stack with hall effect sensors [?]. The stack is made up of a the bottom layer full of rigidly connected three dimensional hall effect sensors, the second layer is made from an elastomer, and the top layer has a magnetic particle unit placed at a set distance above each of the hall effect sensors. The movement of the magnets alters the magnitude and direction of magnetic field sensed and data can be interpolated to create a map of strain deformation. The main advantages of this method is that each hall sensor can detect in three dimensions, hence normal and shear forces can be detected, and using magnetismfor sensing means less electrical noise in the system. The main disadvantages of this method of sensing is the added complexity in scaling the system and the electronics required and the rigid surface required.

2.2.1.4 Optical

There are various methods for making a optically driven artificial skins. A recent review has been curated by Lee et al. [?] all of the different methods of using optics for creating tactile sensors. The main advantages of optical sensors include the high speed sensor response, immunity to electrical noise, and their non-invasive nature. The main disadvantages include, the bulky hardware required for driving the optics and signal processing, the potential interference of external light sources, and the materials that can carry optical signals.

2.2.1.5 Acoustic

Acoustic soft tactile sensing has not been explored much compared to the other forms of sensing given. Park et al., Hughes and Correll [? ?] have created a system which uses passive acoustic tomogrpahy (PAT) to localise and and classify different types of touch. This form of tactile sensing is the most similar to the biological system of mechanoreceptors which are specialised to detect certain frequencies of vibration.

2.2.1.6 Soft Pressure mapping technology comparison

The softness of biological human skin has a large range as discussed in Section 2.1.2. There have been a range of works investigating sensors with a range of softness' and performance. A comparison of these start-of-the-art soft sensor works is given in Table 2.2.

Table 2.2: Comparison of soft sensor technologies.

1st Author	**Sensing principle**	**Sensing region material**	**Sensing region elastic modulus or shore hardness**	**Electrodes per sensing position**	**Repeatability**	**Time series data shown**	**Spatial resolution**	**Temporal resolution**
Gilanizadehdizaj [?]	Piezo-resistive	Ecoflex30-00 rGO composite sponge	40 kPa	2 sensels / electrode	10 cycles for each stress	-	10 x 10 mm	-
Fu[?]	Piezo-resistive	Carbon black silicone composite	1.5 Mpa	0.625 sensels / electrode	50000 cycles	Yes.	12 x 12 mm	60 ms
Yang[?]	Piezo-resistive	Ecoflex graphene composite sponge	-	2 sensels / electrode	800 cycles	Yes.	10 x 10 mm	150 ms
Liang[?]	Capacitive	PDMS, PET, Si, Sio2, Cu laminate	4000 Mpa	1 sensel / electrode	-	Real-time use of sensor shown. No explicit time-series data.	4 x 4 mm	-
Yan[?]	Magnetic	Ecoflex 00-50	83 kPa	11 IC pins / sensel	30,000 cycles	Yes.	0.1 x 0.1 mm	15 ms
Rossiter[?]	Optical	Polymer foam	-	2 sensels / electrode	-	-	10 x 10 mm	-
Shimdera[?]	Optical	Super clear silcone	40 A	N/A. One fiber optic LASER and one camera.	Error increase of 1.7% over 30 days	Yes.	approx. 20 x 20 mm / 0 - 1100 um	Sample rate 1.6s. Training required.
Ramuz[?]	Optical	PDMS	-	N/A. Two arrays of OLEDs and Organic Photo Detectors used.	900 cycles	Yes.	Not localised.	300 ms

2.3 Biological Muscle form and function

Note: This section was taken from literature reviews from 3 years ago, when I was going to research DEAs. Needs a re-review ASAP.

Biological muscles are a product of millions of years of evolution and the motion and other mechanical characteristics of biological structures is yet to be outperformed by artificial muscle technology. To determine how to quantify the performance of a biological muscle this section gives foundational knowledge about muscle function, structure, and how it can be characterised from an engineering perspective rather than the typical biological perspective, so that similar actuator devices can then be investigated which have similar attributes.

Biological muscle is a naturally occurring tissue comprised of muscle fibres bundled together to apply a contractile force on connecting tissue or, in the case of smooth muscle, applying a force on itself. The base actuator units of muscle are proteins myosin and actin filaments, which effectively slide against each other to produce a contractile motion. The root cause of a muscle contraction is an electrochemical signal sent form the central nervous system to a motor neuron/s which travel to the muscle where electrochemical reactions take place for the contraction to occur. The sliding motion of the myosin and actin filaments is due myosin heads

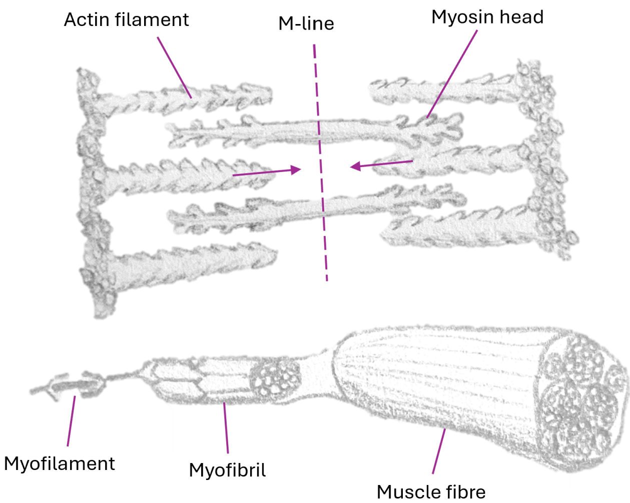

FIGURE 2.5: Components of a biological muscle contractile unit and meta-structure.

attaching to the actin and pulling the actin towards a middle line (M-line) in multiple stroke actions. These filament actuators are stacked in three dimensions within a muscle fibre to amplify contractile stress and strain as shown in Figure **??**.

From a high level, muscle is made up of bundles of fasicles connected together with a tissue called perimysium. Within the fasicles are many muscle fibres (i.e. muscle cells) which are surrounded by a connective tissue called endomysium. Within the muscle fibres there are many sacromeres stacked within a cyclindrical-like structure called a myofibril. Each sacromere contains a contractile unit of myofilaments.

2.3.1 Characterising a muscle

To be able to quantify the performance of a biological muscle there must be certain metrics characterising muscles such that both artificial and biological that can be compared. An artificial muscle can be characterised using typical mechanical material parameters such as:

1. Stress - Force that is applied to the normal of the cross section of the muscle through various states of muscle excitation. $[N.m^{-2} or Pa]$
2. Strain - The muscle change of length due to the stress applied through various states of muscle excitation.
3. Elastic modulus - The elasticity determining the relationship between stress and strain for the linear region of the stress strain characteristic curve. $[Pa]$
4. Energy density - The work done by the muscle per unit volume or mass. $[J.kg^{-1}]$
5. Power density - The work done by the muscle per unit volume or mass per unit time. $[W.kg^{-1}]$
6. Ultimate tensile strength - The maximum tensile stress that a material can tolerate before breaking. $[Pa]$
7. Efficiency - The work done by the muscle compared to the energy put into the system, known as metabolic cost in biological muscles. [%]
8. Actuation frequency - The frequency range of actuation cycles using the system's method of excitation. $[Hz]$
9. Stroke - The maximum displacement an actuator can achieve $[m]$
10. Life cycle - The time or number of actuation cycles in which it takes for the actuator to degrade such that it cannot perform its intended purpose to specified standards.

    As well as the commonly used medical/biological muscle metric:

11. Maximum isometric contraction force - the maximum force a muscle can apply without changing strain. This is also related to the ratchet-like mechanism and muscle locking where a muscle can apply a much larger force in a static state, as seen in the myosin binding[**?**].

Other qualities of muscle should be quantified on a case by case basis depending on the artificial muscle technology being investigated. For example, a major issue with dielectric elastomer actuators is the excitation voltage required for actuation is too large for many applications. Hence this could be another parameter considered for some artificial muscles.

Some of the biological muscle metrics have been quantified by previous research as seen below:

- Energy density - energy densities ranging from 0.4 - 40 $J.kg^{-1}$[**?**].
- Power density - power densities ranging from 9 - 284$W.kg^{-1}$[**?**]
- Actuation frequency - natural actuation frequencies ranges 1 to 180Hz[**?**].
- Strain - ranging from 5 - 30%[**?**].
- Efficiency - Thermodynamic efficiency of human muscle is typically between 20-35%[**?**]. However other biological muscle has been seen to reach efficiencies of up to 77%[**?**].

2.3.2 Muscle Mechanics

Before attempting to recreate a bio-mimetic actuator it is important to acknowledge the numerous simplified electro-mechanical system models of parts of the muscle actuation process. These models need to be understood to gain an understanding of the application of biomimetic actuators can be used in assistive soft robotic devices. From here we will present basics of the subject of bio-mechanics.

The stress and strain involved in muscle contraction is more complex than uniform materials and is non-linear. The stress and strain of a passive muscle (i.e. contractile units are not producing internal muscle tension) can be modelled with the following equation;

$$\frac{d\sigma}{d\varepsilon} = \alpha.(\sigma + \beta) \tag{2.1}$$

Where ε & σ are strain and stress respectively. A solution for this is first order ODE is;

$$\sigma = \mu e^{\alpha\varepsilon} - \beta \tag{2.2}$$

Where μ is a free parameter determined empirically. The stress-strain of a passive muscle can be likened to tension being applied yarn. As more strands of the yarn are pulled into tension the stress increases, then as the last strands are brought into tension a maximum stress is reached, until the yield stress is reached. Linear approximations can still be made over regions of elongation depending on accuracy required for application. The stress-strain of an active muscle (i.e. when it is tetanised) is approximated to a piece-wise quadratic function or bell curve. It is important to note that the stress for both active and passive muscle is zero when the strain is less than 0.4, demonstrating the yarn-like nature of the muscle stress-strain. Hill's muscle models commonly refer to a mechanical three element model [**?**] composed from, one parallel non-linear spring element, one series non-linear spring element, and a contractile unit.

2.3.3 Electrical Muscle Models

Similar to EAP-based artificial skin and artificial muscles, biological muscles also require electrical stimulation to function. The main method for providing an artificial electrical stimulation to a muscle, to simulate the signal a motor neuron would give to a muscle, is functional electrical stimulation (FES). Due to the biochemical nature of the motor neuron signal transport and the purely electrical stimulation provided by the FES device, the process isn't as efficient as the naturally occurring electro-chemical muscle activation, often resulting in increased muscle fatigue when compared to equivalent voluntary muscle contractions [**?**]. FES applies a voltage across between two electrodes on the user's skin above a specific muscle. The voltage simulates the signal form and frequency of action potentials between 4 - 12Hz[**?**]. The threshold for a muscle action potential to cause a muscle contraction is approximately 70 mV [**?**]. EMG also commonly uses two electrodes on the surface of the skin above a desired muscle. EMG senses the nerve impulses sent to the muscle and propagated through action potential.

2.4 Artificial Muscle Technology

There are many types of electrically actuated artificial muscles technology. Artificial muscle actuator technology that has gained particular interest in recent years include, the ionic polymer-metal composite (IPMC) actuator, the hydraulically amplified self-healing electrostatic

(HASEL) actuator, magnetorheolgical elastomer (MRE) actuators, and dielectric elastomer actuators (DEAs). Each of these having qualities very similar to that of biological muscle usually with a trade-off in actuation response time, actuation force, and actuation strain for their various possible topologies. This section gives a brief overview of four state-of-art soft electromagnetically driven actuator technologies.

2.4.0.1 Ionic polymer–metal composite actuator

Ionic polymer-metal composite actuators (IPMCs) are soft actuators that can be actuated at a much lower excitation voltage than DEAs, commonly being less than 10V. IPMCs are also desirable as artificial muscles they have shown large bending deformations, simple to fabricate, light weight and thin in design, and can have a fast actuation response time (>15Hz) at small displacements[**?**]. IPMCs also have a high work density and maintain a constant volume during actuation like biological muscles[**?**]. An IPMC is made up of an ionic polymer interlayer, two

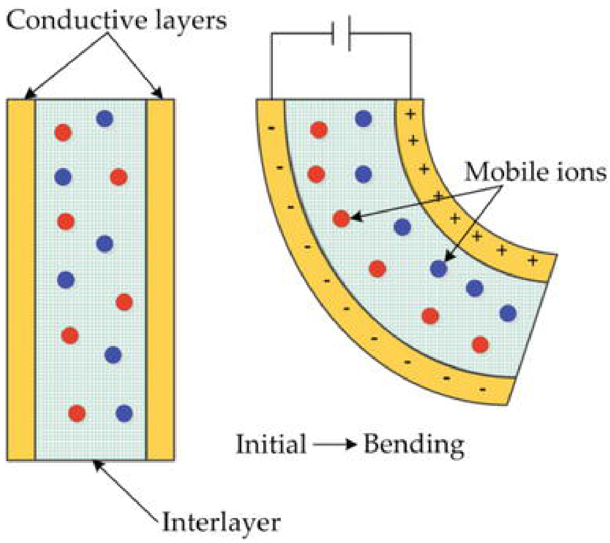

Figure 2.6: Diagram of the typical architecture of an IPMC actuator[**?**] (© 2018 Yanjie Wang and Takushi Sugino)

electrode conductive layers, and a voltage source. The ionic polymer interlayer allows for ionic transport and is typically made of treated Nafion or Flemion. These materials are typically used as ion exchange membranes so have the characteristics desired for the transporting ions during the actuation of the IPMC actuator. The two electrodes are made of a suitably conductive and flexible material. The interlayer is treated such that it is filled with water molecules and cations, with the chemical backbone of the interlayer being slightly negatively charged. When a voltage is applied across the electrodes the cations are repelled from the cathode and travel towards the anode while the water molecules are displaced in the opposite direction towards the cathode. The ionic polymer then swells as the cations repel each other along the anode side of the interlayer, while the polymer elements on the cathode side effectively shrink[**?**]. This swelling adjacent to the cathode provides the device's bending actuation.

There are many variations of the design and manufacturing of IPMCs to optimise the actuator for an application as shown by [**?**].Although the process of manufacturing IPMCs is simple, it takes a long amount of time (often can be over 48hours[**?**]) for the ionic polymer interlayer to absorb the necessary ions and undergo the necessary reactions. There has been much research into the optimal manufacturing of an IPMC [**? ? ?**]. The use of additive manufacturing has been used successfully to generate more complex geometries using fused filament deposition[?].

IPMCs can also be used as sensors. When an IPMC undergoes bending due to an external force there is a potential generated across the electrodes, which indicates bending direction and magnitude[?].

Two key deficiencies of current IPMC actuator technology are the maximum force output achievable and the life cycle of the actuator in a dry (non-aqueous) environment.The force output optimisation of IPMCs has been investigated by several researchers, all of which having a maximum actuation force in the milli-newton scale [? ? ?]. Because the IPMC actuators rely on hydrated ionic transport to actuate this means if the IPMCs are in a dry environment then over time they will decrease their maximum actuation force.

The applications of this actuator is limited to applications requiring a small actuation force and a wet environment. Current applications include flexible catheters [?], small biomimetic robotics [? ?], aquatic robotics[? ?], with many other applications yet to be discovered.

2.4.0.2 HASEL actuator

A hydraulically amplified self-healing electrostatic (HASEL) actuator is a recent soft actuator technology developed in 2018[?] which displays many qualities that are better than current artificial muscle technology. HASEL actuators are made up of three main components: electrodes, dielectric fluid, and an elastomeric shell. The electrodes need to be highly conductive, able to handle high electric potential, and can be solid or flexible. Hydrogel electrodes have been proven to be a good material for the electrodes because of their elasticity while still maintaining a high conductivity[?]. In one application the hydrogel material is bonded to a polydimethylsiloxane (PDMS) substrate for mechanical strength and for ease of bonding to the actuator biaxially-oriented polypropylene (BOPP) shell[? ?]. HASEL actuators use high electric potential across two electrodes to create an electrostatic force. This force induces a 'zipping' effect which pulls the electrode together from one end to the other as the electric field strength increases. The zipping of the two electrodes pushes the dielectric fluid into the reservoir increasing the pressure which alters the shape of the reservoir bounds providing an actuation motion. When the electrodes have displaced all of the fluid between them the actuation displacement is at a maximum. The electrostatic zipping action allows a large force to be generated due to snap-through transition. Snap-through transition is an actuation instability which has been discussed in previous research as a means of amplifying DEA actuation strain[?]. Recorded efficiency values of HASEL actuators of 21% are comparable to that of human muscles of 20 - 35% [?]. The actuators have had a frequency response of up to 20Hz. Large strains of 124% have been recorded, but can only be achieved when actuating at a resonant frequency. Strains of up to 79% have been recorded using a linear planar HASEL actuator configuration and DC voltage stepping. Else, strains of only 10% have been recorded for static steady strain[?]. Because there is a relationship between the motion of the actuation and capacitance between the electrodes, this means self sensing can be achieved through the electrodes. Although due to the flexible and fluid nature of the device, modelling of the HASEL is difficult and limited in accuracy.

The simple and commonly used manufacturing process for HASEL actuators is completed in six steps as shown by the diagram below:

Other attempts have been made to use polyjet inkjet based additive manufacturing to make the whole HASEL actuator and have been successful with proof of concept, but are yet to be developed from prototype stage[?].

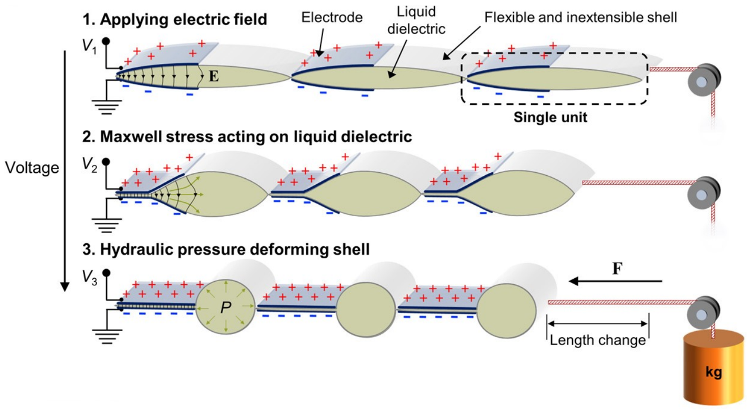

FIGURE 2.7: Diagram of the typical architecture and the contraction stages of a HASEL actuator[?]

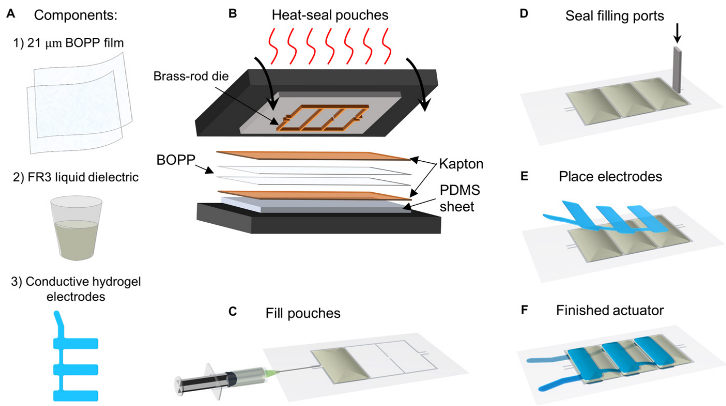

FIGURE 2.8: Diagram of the simplified stages of HASEL actuator production[?]

The cyclic life of HASEL actuators are high, because of their 'self-healing' properties. When there is a dielectric breakdown through the liquid dielectric the damage caused is not permanent like when a DE breaks down. The liquid may form some small air bubbles, however these may not effect the operation of the actuator, instead this can increase the likelihood of a another dielectric breakdown. The cycle life of the HASEL actuator was seen to be larger than one million with a given torus shaped HASEL actuator[?].

The number topologies possible with HASEL actuators is vast. Some topologies of HASELs include torus, planar linear[?], scorpion metasoma[?].

2.4.0.3 Magnetorheological Elastomer

Magnetorheological elastomer (MRE) actuators are a relatively new form of actuator however the theory reinforcing operating principle has been known since at least the 1980s [?]. The structure of an MRE actuator generally consists of a ferromagnetic elastic composite and a driving magnetic field. An example of this is a composite of iron-carbonyl powder and PDMS. The The operating principle of these are that magnetic flux travelling through the the MRE will change mechanical characteristics within the elastomer (i.e. stiffness or displacement of the body). The operation of a MRE actuator is similar to a DEA however instead of having an electric field cause a contraction it is a magnetic field causing a deformation. An MRE is typically made of silicone rubber containing magnetic ferrite based particles uniformly distributing throughout its volume. This kind of actuator is current controlled and can hence operate at a low voltage. This helps mitigate the risk of electric shock of a device in close proximity to humans (unlike HASEL actuators and DEAs).

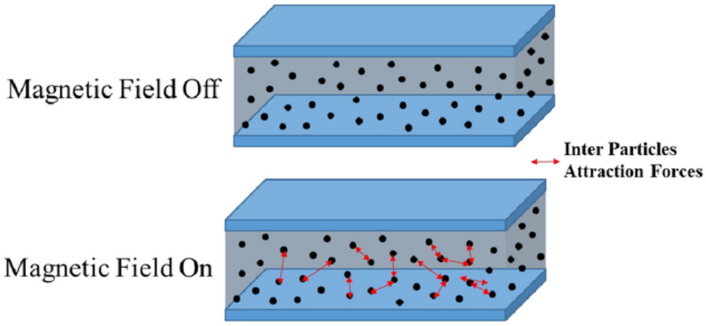

FIGURE 2.9: Diagram showing MRE contraction actuation when a magnetic field is applied[?]

A key issue with using magnetorheological elastomers as soft actuators is that they require heavy gauge conductors for the high current they require for generating a magnetic field. The high current requirement means that actuators have only been created that have a solid electromagnet driving a soft MRE[?].

When manufacturing MREs, uncured liquid silicone rubber is mixed with magnetic (commonly carbonyl iron) particles to form a 3 dimensional matrix of crosslinks with the magnetic particles fixed between the crosslinked polymers. A key issue when creating an MRE is the conglomeration of magnetic particles due to residual water within the mixing operation. The magnetic particles can be processed to have a hydrophobic quality to mitigate this issue. During the curing process a magnetic field can be applied to align the particles within the elastomer as it becomes more rigid.

There have been attempts to use additive manufacturing to make MREs[?], however the method described has not optimsied the structure of MRE for any application and the dipersion of MRE is not uniform throughout the print volume.

The current applications of MRE actuators are limited, however magnetorheological fluid (MRF), is a fluid which becomes more viscous with an applied magnetic field as currently has many modern applications. This fluid substance is largely used in applications where damping control is desired such as vehicle suspension[?], medical assitive devices[?] and helicopter seat damping [?]. Potential MRE actuator applications include fluid valve control[?] and active vibration control similar to that mentioned for MRFs[?].

2.4.0.4 Dielectric Elastomer Actuators

The dielectric elastomer actuator (DEAs) are often called artificial muscles because they share similar characteristics to biological muscle such as, the large strains achievable, the high elastic energy density, many topologies/configurations achievable, and constant volume during its contraction.

A DEA consists of a dielectric elastomer (DE) film sandwiched between two compliant electrodes. To excite the actuation, a high electric potential is applied to across the electrodes creating an electrostatic force between the two compliant electrodes. This force pulls the two electrodes together applying stress (known as Maxwell's stress) to the elastomer and hence strain parallel and perpendicular to direction of the electrostatic force. When the DEA is contracted the surface area of the electrodes increases and the thickness of the DE decreases causing a change in capacitance and Maxwell's stress. A dielectric elastomer actuator can be modelled

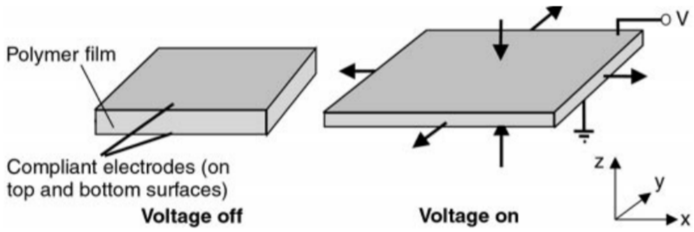

FIGURE 2.10: Diagram of a DEA with a with no voltage and a voltage applied across the electrodes. [?]

as a flexible parallel plate capacitor in its simplest form. Using this we can determine the electrostatic pressure to be:

$$\sigma_{es} = \varepsilon_0 \varepsilon_r \frac{V^2}{z^2} \tag{2.3}$$

Where p_{ES} is the electrostatic pressure, ε_0 and ε_r are the vacuum and relative permittivity constants, V is the voltage potential applied across the electrodes and z is the thickness of the DE. The electrodes used for a DEA need to be made of a conductive material, but require similar elasticity to the dielectric material. An ideal material for these electrodes would have high conductivity. This conductivity would change minimally and predicatively under large strains. Many composites have been used in practice for these electrodes, with the most common in early development being a silicone rubber and carbon powder composite. However, the unpredictable nature of carbon powder elastomer composites has lead to research into many other materials/silicone additives such as hydrogels, graphene sheets, metallic nanostructures, carbon nanotubes, liquid metal[? ? ? ?]. The ideal material for the dielectric elastomer should have a high elastic modulus and a high electric breakdown voltage. The elastic modulus needs to be sufficiently high so that less electrostatic pressure can create a larger strain. While the breakdown voltage of the material needs to be sufficiently high such that the material will not break down at the maximum desired strain. If a material can be found with a high enough electric breakdown strength at a smaller thickness than current research prototypes then a higher stress can be achieved giving a larger or equivalent actuation force at a lower voltage.

Many other topologies exist to generate different actuation motions using the same electrostatic pressure generation principle. These include actuator topologies such as stack[? ?], helical[?], bending[?], lens[?], cylindrical, and rolled shaped actuators[?]. Each of which having a range of applications.

DEAs are often fabricated in a laboratory environment using a pre-strained elastomer. The pre-straining does three key things; stores elastic strain energy, ensures DE is planar within the bounds of the jig, and controls the initial thickness of the elastomer. There is no standard practice for the fabrication of DEAs, other methods such as additive manufacturing have also been explored to generate more complex geometries and to increase production speed[? ?].

As well as actuating, DEAs can also be used for sensing. DEAs can be used as sensitive capactive sensors, where any strain applied to the DE will relate to the effective capacitance between the two electrodes[? ? ?].

Currently dielectric elastomer actuators all require voltages within the kilo-volt range to generate what can be called a useful stress and strain for many applications. A key problem encountered by researchers designing DEAs is the trade-off between actuation force and strain magnitude [**?**]. This high voltage requirement makes the technology dangerous for use where there is a possibility that a human may come into physical contact with the high voltage electrodes.

2.5 Soft Conductive Particle Piezoresistive Composites

Soft sensors and actuators require low-stiffness materials for their active sensing/actuation domains. The requirement of softness is governed by the mechanical modulus values depend on the application requirements. The use of conductive particle elastomer composites is explored in this work due to the customisability of the electromechanical characteristics.

A core part of this work is understanding the behaviour of conductive particle elastomer composites beause of their use as EAPs which can be used for a range of sensing and actuating purposes. The characteristics that make conductive particle elastomer composites (CPECs) ideal for soft sensor and actuator devices often include:

- Low stiffness
- Changeable conductivity
- Piezoresistivity
- Mouldable
- 3D printable
- Low toxicity
- Durable
- Inexpensive
- Easy to obtain
- Simple fabrication process
- Sustainable

2.5.1 Fabricating Conductive Particle Elastomer Composites

Before exploring the known conduction and piezoresistive mechanisms and models for CPECs, it is important to understand how the fabrication process of a CPEC may affect its physical structure.

CPECs are made by dispersing conductive particles through a curable liquid elastomer matrix. To change the electromechanical properties of the material, the dispersion of the conductive particles throughout the matrix can be optimised through various methods. To minimise the agglomerations of primary conductive particles often a sonication step is completed. This involves a mixture of the conductive particles and a liquid, usually in the form of a solvent, to be placed in a sonication bath. The sonication bath performs a frequency sweep whereby the resonant modes of the agglomerates are met causing separation of the agglomerates into their primary particles []. The degree of dispersion is governed by the time in the sonication bath, the sonication frequencies, and sonication amplitudes []. This sonication usually occurs before the the particles are added to the elastomeric matrix due to the large viscous damping effects of liquid elastomers. The next step involves mixing the dispersed conductive particles throughout the liquid elastomer, this can be done using a variety of mixing methods, including a planetary mixer, magnetic mixer, screw mixer, static mixers, amongst others []. During the mixing process often the liquid solvent used in the dispersion stage is evaporated, leaving only the curable elastomer and the conductive particles. When sufficient mixing of the liquid elastomer and conductive particles have been completed the material is formed into a desired final shape using advanced additive manufacturing methods [] or traditional moulding [] or film making techniques []. During the moulding process the material undergoes a form of curing, such as UV curing, catalysed curing, or moisture curing[]. If the composite material has not already been integrated into a device containing electrodes and other mechanical support structures these are integrated at the end of the process [].

2.5.2 Modelling Conduction mechanism

The typical fabrication process stated in Section 2.5.1 for CPECs shows that the dispersion of conductive particles will always vary.

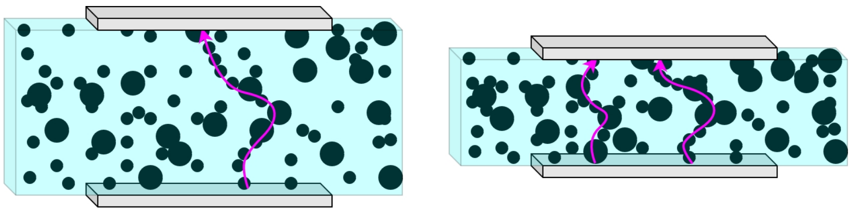

FIGURE 2.11: Two grey electrodes across a CPEC cuboid showing enlarged black conductive particles within a blue polymer matrix. Left: An uncompressed CPEC. Right: A compressed CPEC.

Some of the physical features of these conductive percolation networks can be quantified and directly relate to the macro-level electromechanical properties of the material. Such characteristics of a conductive percolation network include:

1. Conductive particle(s) used
    - (a) Aspect ratio [**?** **?**]
    - (b) Inherent particle conductivity
2. Conductive particle dispersion [**?**]
    - (a) Inter-particle distance distribution
    - (b) Particle agglomeration distribution [**?**]
    - (c) Isotropy/anisotropy [**?**]
    - (d) Sedimentation [**?**]
3. Elastomeric matrix
    - (a) Viscosity
    - (b) Elastic modulus
    - (c) Dielectric permittivity

4. Impurities
5. Voids

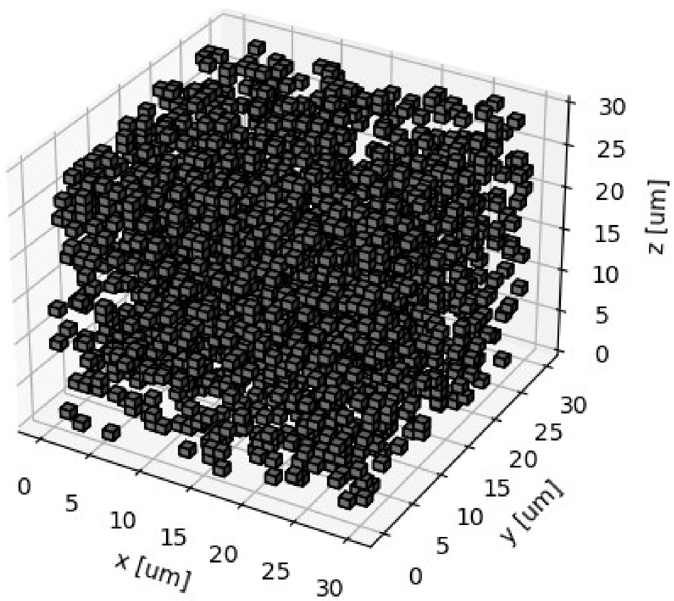

FIGURE 2.12: Example of a randomised cube percolation with a volume percentage of 8% of particles

Microscale models for CPECs and the relationship between particle and electric charge motion are often computationally heavy, overly idealised, and non-invertible [?]. A microscale model example can be seen in Figure 2.12. However, microscale modelling of CPECs may give insight into understanding complex physical phenomena that may relate to the macroscale models made for CPECs. An alternate method for modelling CPECs is the formation of macroscale models[?].

Electrical DC conduction through a CPEC occurs using two main mechanisms, Coulomb conduction and quantum tunneling [? ? ? ?]. Coulomb conduction uses the conduction band electrons are shared by adjacent atoms allow conduction throughout chains of cascading conductive particles. The second mechanism of conduction is through quantum tunneling which is stochastic in nature and allows for conduction through insulative boundaries between the percolative network of conductive particles [? ?].

Electrical AC conduction can occur through a CPEC through capacitive means depending of particle spacing with a decrease in reactance becoming more prominent for composites near the percolation threshold[?].