Soft Electroactive Elastomer Bodies that Can Sense and Contract

by

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in

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Declaration of Authorship

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“Today.“

R. Ellingham, August 2024

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Abstract

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The Thesis Abstract is written here (and usually kept to just this page). The page is kept centered vertically so can expand into the blank space above the title too…

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Abbreviations

EIT	**E**lectrical **I**mpedance **T**omography
DEA	**D**ielectric **E**lastomer **A**ctuator
LAH	**L**ist **A**bbreviations **H**ere
LAH	**L**ist **A**bbreviations **H**ere
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Physical Constants

Speed of Light $c = 2.997\ 924\ 58 \times 10^{8}\ \mathrm{ms}^{-\mathrm{S}}$ (exact)

Symbols

Sensels	A single sensor element of an array of sensor
F	force
add	here...

Dedicated to baked beans in all their glory...

Chapter 1

Introduction

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 tools for creating and characterising artificial pressure sensitive skin technology and then continues to explore the integration of this artificial skin technology into an artificial muscle technology.

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[23–27], sensors[28], and soft robotics control[29, 30]. 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[25]. 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[31–34].

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)[35–38] and ionic polymer-metal composites (IPMCs)[39]. 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. Use additive manufacturing methods and design a mixer for FDM printing of conductive particle composites for soft sensors and actuators.
3. From the characterisation in objective 1 mitigate the effects of the transient phenomena.
4. Create a set of metrics for quantifying the performance of an electrical impedance tomography based artificial skin.
5. Simulate and integrate an electrical impedance tomography based artificial skin onto a dielectric elastomer actuator.

1.3 Chapter Contributions

Chapters 3 - 7 contain the core novel research contributions. Chapters 2 and 8 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 - A Novel Mixing Method for 3D Printing Conductive Particle Elastomer Composites: This chapter discusses the place for advanced manufacturing for 3D printing for soft actuator and sensor technology and a new form of mixing for such manufacture.

Chapter 5 - 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 6 - 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 7 - Hardware for a DEA-EIT Sensor Actuator Hybrid Device:: 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: 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.

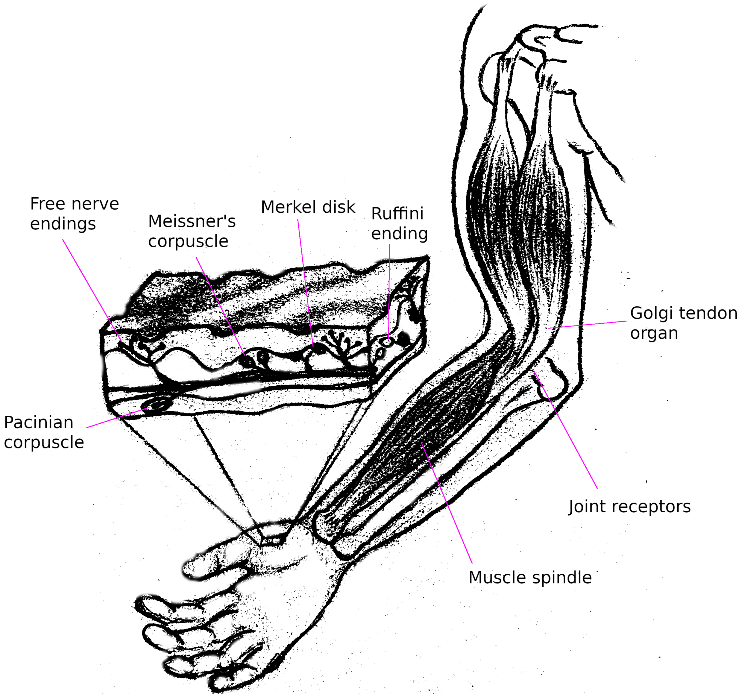


FIGURE 2.1: Examples of the locations of proprioceptors and cutaneous mechanoreceptors in the human body.

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[1–4]. Examples of such devices are displayed in Figure 2.2

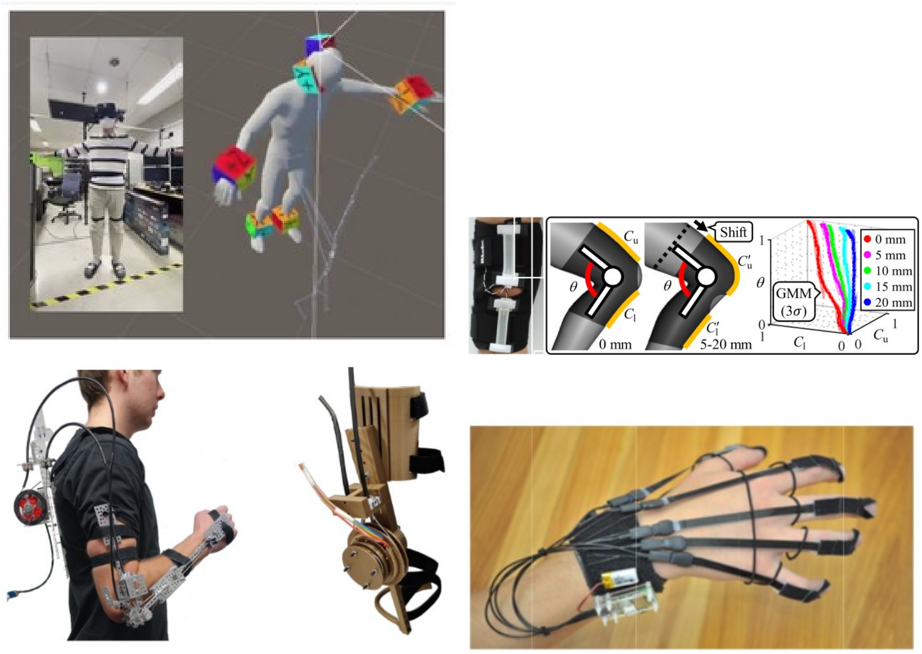


FIGURE 2.2: Clockwise from top left: IMU pose estimation[1], stretch sensor knee joint pose estimation[2], stretch sensor hand joint pose estimations[3], encoder elbow pose joint estimation[4].

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[5], Pressure Profile Systems pressure sensors on a robotic hand[6], Soft pressure mapping gripper[7], Tekscan thin pressure mapping platform[8], Tactilus seat pressure mapping system[9]

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 [22].

	A - Glabrous skin			B- Both	
Receptor	**Meissner corpuscle** *A1*	**Ruffini corpuscle** *A2*	**Pacinian corpuscle** *A3*	**Merkel cell-neurite complex** *B1*	**Free nerve ending** *B2*
Perceptual sensory functions	Skin movement, handing objects	Skin stretch, movement direction, hand shape and finger position	Vibration when grasping an object	Fine tactile discrimination, form and texture perception	Pain, nociception
Skin stimulus	Dynamic deformation	Skin stretch	Vibration	Indentation depth	Injurious forces
Localization	Dermal papillae	Dermis	Deeper dermis	Basal layer of the epidermis / around guard hair	Epidermis > Dermis
Afferent response / Stimulus	RAI-LTMR	SAII-LTMR	RAII-LTMR	SAI-LTMR	SA-HTMR
Associated fibre	Aβ	Aβ	Aβ	Aβ	Aδ C-HTMR
Conduction velocity	35-70 m/s	35-70 m/s	35-70 m/s	35-70 m/s	5-30 m/s 0.5-2 m/s
Receptive field	22 mm^2	60 mm^2	finger, hand	9 mm^2	1-3 mm^2
Receptor / Hair density	150 / cm^2	10 / cm^2	20 / cm^2	100 / cm^2	4-25 / mm
Putative MS ion channels	β-ENaC/γ-ENaC/ASIC2/ASIC3/TRPV4/KCNQ4/Piezo 2	β-ENaC/ASIC3/Aquaporine 1/Piezo 2	β-ENaC/γ-ENaC/ASIC1/ASIC2/Piezo 2	ASIC2/ASIC3/TRPV4/Piezo 2	TRPA1/TRPV1/TRPV2/TRPV4/Piezo 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. Shear modulus - The relationship between the shear stress and shear strain in the linear region of the stress-strain characteristic curve [Pa]
4. Ultimate tensile stress (UTS) - The maximum tensile stress that a material can tolerate before breaking [Pa]
5. 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

6. 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]
7. Skin thicknesses - the thickness of all layers of skin the cutaneous epidermis and dermis and thickness of the hypodermis [mm]
8. Skin surface area - Biological skin has a large surface area and can also be regionalised to map skin function and sensitivity [m^2]
9. 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 [41–43]
2. Static force resolution - This is the detection resolution of static or slow-acting forces acting upon the skin [43]
3. Temporal resolution - This is the detection resolution of fast-acting forces acting upon the skin often required for texture recognition [41, 43]

A numerical 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 [44], 0.1 - 2.4 MPa [45], and 10.4 - 89.4 kPa [46].
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 [47], 473.9 $\pm$ 42.5 Pa and 32.3 $\pm$ 10.0 Pa at 205 Hz [48].
3. Shear modulus - Shear modulus has been reported to be 100 times that of elastic modulus for upper most layers of skin (epidermis and stratum corneum) [49]
4. Ultimate tensile stress - 21.6 $\pm$ 8.4 MPa [44]. 28.0 $\pm$ 5.7 MPa [50]
5. Life cycle - Skin cells are constantly growing, dying, and shedding. Skin is always actively remodelling based on external stimuli.
6. Strain creep - The strain creep was found to be 2.7 kPa.s for a 10 Pa step input on a dermis skin sample [47].
7. Stress relaxation -
8. Skin thicknesses - The thickness of human cutaneous skin ranges from 0.6 to 2.6 mm with an average skin thickness of 2 mm [41].

9. Skin surface area - The average surface area of skin in adult humans is 1.7 ± 0.1 m^2 [41].

10. Isotropy/Anisotropy - The tension lines in skin are determined by collagen fibre orientation and dynamic stretch events [51, 52]. The elastic modulus of human skin was reported to be 160.8 ± 53.2 MPa parallel to the skin tension lines and 70.6 ± 59.5 MPa perpendicular to the tension lines [50]. The UTS of human skin was reported to be 28.0 ± 5.7 MPa parallel to the tension lines and 15.6 ± 5.2 MPa perpendicular to the tension lines [50].

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 [53]. Two point discrimination is another metric for determining spatial resolution an has been determined as 3.7 ± 0.7 mm [54]. 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 [22].

2. Force resolution - Minimum force detection on various regions of human skin was found to be between 67 - 1007 mg [55], and various mechanoreceptors 0.73 - 122.6 mN [56].

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 [53]

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.[41] 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 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.

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 take place. The sliding motion of the myosin

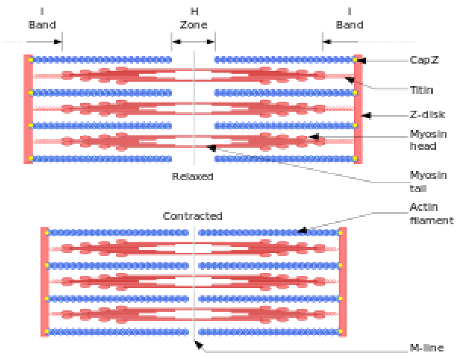


FIGURE 2.4: Components of a biological muscle contractile unit [10]

and actin filaments is due myosin heads 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 2.4.

The anatomy of a human skeletal muscle can be seen in Figure 2.5. The 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.

2.2.1 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.

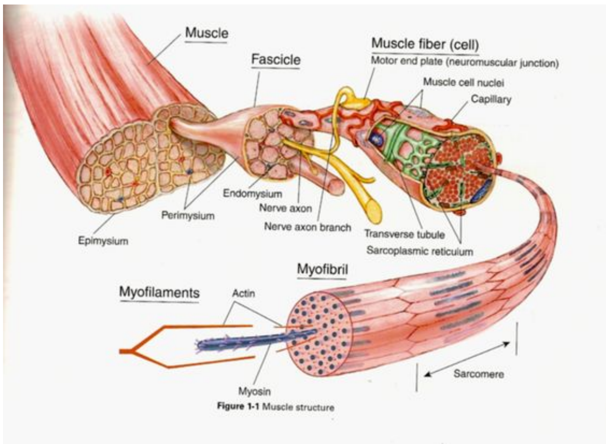


FIGURE 2.5: Diagram of the internal structures of a skeletal muscle[11]

3. Elastic modulus - The elasticity determining the relationship between stress and strain for the linear region of the stress strain characteristic curve. $[Pa]$
4. Shear stress - Force applied parallel with a cross sectional area plane due to a state of muscle excitation. $[N.m^{-2} or Pa]$
5. Shear strain - The change in deformation perpendicular to the direction of loading to the due to a state of muscle excitation.
6. Shear modulus - The relationship between the shear stress and shear strain in the linear region of the stress strain characteristic curve. $[Pa]$
7. Energy density - The work done by the muscle per unit volume or mass. $[J.kg^{-1}]$
8. Power density - The work done by the muscle per unit volume (or mass) per unit time. $[J.kg^{-1}s^{-1} or W.kg^{-1}]$
9. Yield stress - Stress at which the stress strain curve of the muscle begins to become non-linear and the material strain may not return to it's resting (original) length. $[Pa]$
10. Ultimate tensile strength - The maximum tensile stress that a material can tolerate before breaking. $[Pa]$
11. Efficiency - The work done by the muscle compared to the energy put into the system, known as metabolic cost in biological muscles. [%]
12. Actuation frequency - The frequency range of actuation cycles using the system's method of excitation. $[Hz]$
13. Stroke - The maximum displacement an actuator can achieve $[m]$
14. Drift - Change in actuation displacement over time given the same excitation input value each actuation cycle. $[m]$
15. 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 commonly used medical/biology muscle metrics such as:

16. 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[58].

17. Muscle force direction and architecture - Biological muscles can have varying contraction force directions determined by pennation angle of the muscle and the muscle fibre configuration.

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 - Biological muscle can have energy densities ranging from 0.4 - 40 $J.kg^{-1}$[59].
- Power density - Biological muscle can have energy densities ranging from 9 - $284 W.kg^{-1}$[60]
- Actuation frequency - The range of natural actuation frequencies for both vertebrate and invertebrate muscles ranges 1 - $180 Hz$[60].
- Strain - Biological muscle can have strains ranging from 5-30%[61].
- Efficiency - Thermodynamic efficiency of human muscle is typically between 20-35%[62]. However other biological muscle has been seen to reach efficiencies of up to 77%, such as in tortoises[62].

2.2.2 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.

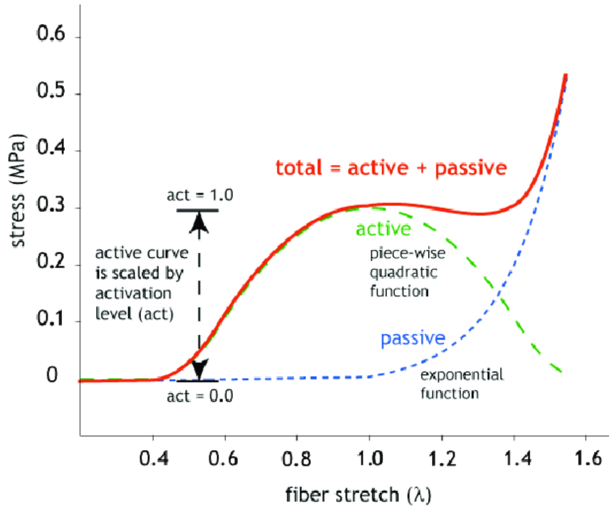


FIGURE 2.6: Plot showing the stress and strain of active and passive muscles [12]

Hill's muscle models commonly refer to a mechanical three element model [63] composed from, one parallel non-linear spring element, one series non-linear spring element, and a contractile unit, displayed as a free-body diagram in Figure 2.7 with the corresponding Equation 2.3.

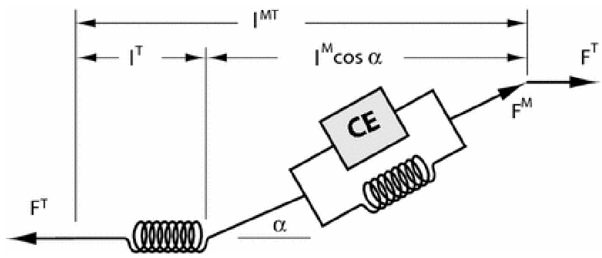


FIGURE 2.7: Hill muscle model[13]

Where F^{KT} and F^{KM} are the spring forces of the tendon and muscle respectively, which are a function of extension length. F^CE is the contractile force and F^T is the total contractile force as observed at the end of each tendon either end of the muscle. Where F^T is the tendon force; F^M is the muscle force; the l^T, l^M, l^{MT} are muscle length, tendon length and their combined lengths respectively; α is the pennation angle (i.e. zero if parallel muscle); The left and right non-linear spring elements represent a tendon and muscle spring characteristic respectively; The CE box represents the contractile element that generates contractile force.

$$F^T = F^{KT} + (F^{CE} + F^{KM})cos(\alpha) \tag{2.3}$$

2.2.3 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 [64]. 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[65]. The threshold for a muscle action potential to cause a muscle contraction is approximately 70 mV [66]. EMG also commonly uses two

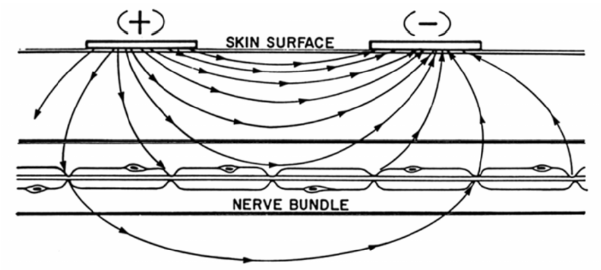


FIGURE 2.8: Electric field generated by two electrodes on the surface of the skin above a specific muscle and hence it's activating nerve bundle[14]

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.

2.2.4 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.

2.2.4.1 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 10V. IPMCs are also desirable as artificial muscles as they can display large bending deformations, simple to fabricate, light weight and thin in design, and can have a fast actuation response time (¿15Hz) at small displacements[67]. IPMCs also have a high work density and maintain a constant volume during actuation like biological muscles. An IPMC is

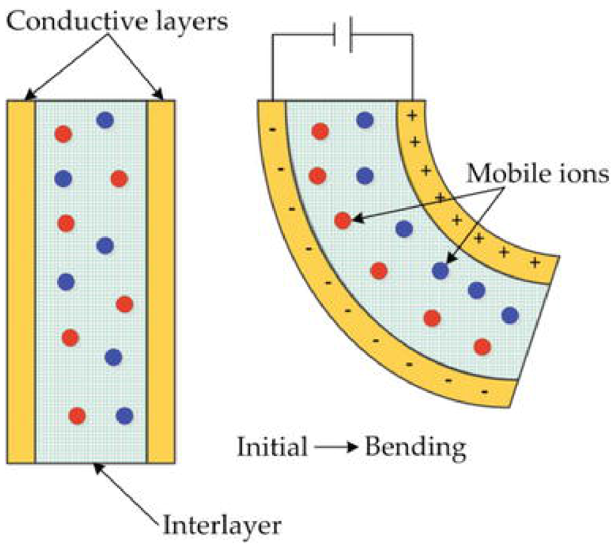


FIGURE 2.9: Diagram of the typical architecture of an IPMC actuator[15]

made up of an ionic polymer interlayer, two 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[68]. 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 [69].Although the process of manufacturing IPMCs is simple, it takes a long amount of time (often can be over 48hours[67]) 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 [69–71]. The use of additive manufacturing has been used successfully to generate more complex geometries using fused filament deposition[72].

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[73].

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 [73–75]. 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 [76], small biomimetic robotics [77, 78], aquatic robotics[79, 80], with many other applications yet to be discovered.

2.2.4.2 HASEL actuator

A hydraulically amplified self-healing electrostatic (HASEL) actuator is a recent soft actuator technology developed in 2018[16] 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[81]. 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[16, 82]. 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[83]. Recorded efficiency values of HASEL actuators of 21% are

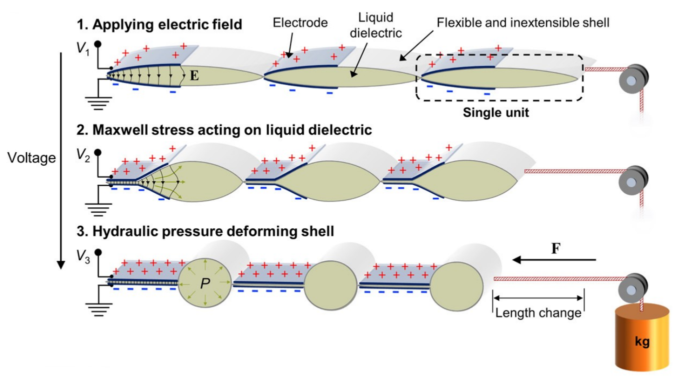


FIGURE 2.10: Diagram of the typical architecture and the contraction stages of a HASEL actuator[16]

comparable to that of human muscles of 20 - 35% [62]. 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[16]. 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:

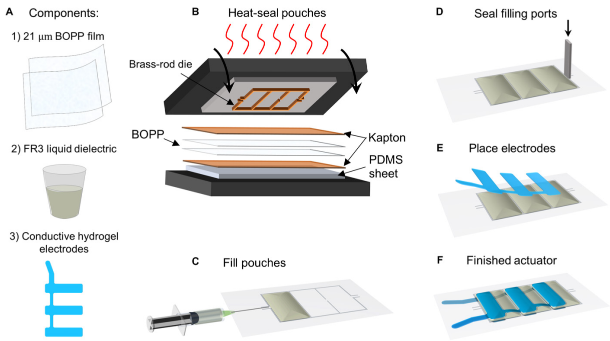


FIGURE 2.11: Diagram of the simplified stages of HASEL actuator production[16]

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[84].

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[81].

The number topologies possible with HASEL actuators is limitless. Some topologies of HASEL actuators include torus, planar linear[81], scorpion metasoma[85].

2.2.4.3 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 [86]. 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).

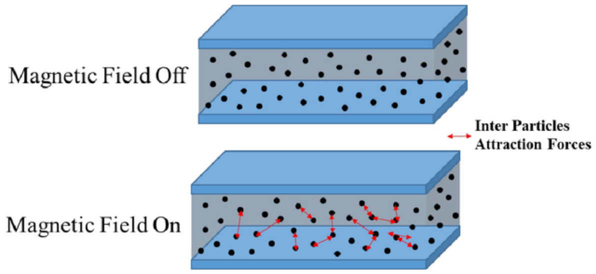


FIGURE 2.12: Diagram showing MRE contraction actuation when a magnetic field is applied[17]

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[87].

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[88], 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[89], medical assitive devices[90] and helicopter seat damping [91]. Potential MRE actuator applications include fluid valve control[87] and active vibration control similar to that mentioned for MRFs[89].

2.2.4.4 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

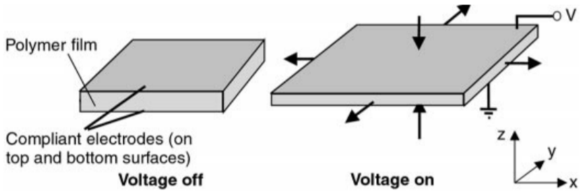


FIGURE 2.13: Diagram of a DEA with a with no voltage and a voltage applied across the electrodes. [18]

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

$$\sigma_{es} = \varepsilon_0 \varepsilon_r \frac{V^2}{z^2} \tag{2.4}$$

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[92–95]. 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[96, 97], helical[98], bending[99], lens[100], cylindrical, and rolled shaped actuators[101]. 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[102, 103].

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[35, 104, 105].

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 [96]. 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.

2.3 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.

2.3.1 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.

2.3.1.1 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 [106–108]
- Intrinsically conductive polymers [107, 109]
- Microfluidic metals [110–112]
- Hydrogel structures [82, 113, 114]
- Flexible piezoresistive semiconductors [115, 116]

	Conductivity	Piezo-resistivity	Change stiffness	Fabrication	Cost	Environmental Stability	Toxicity
Intrinsically conducting polymers	- Dependent on polymer used. (S/cm [109])	- 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.

2.3.1.2 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 relies on the deformation of the capacitor dielectric and/or the capacitor electrodes.

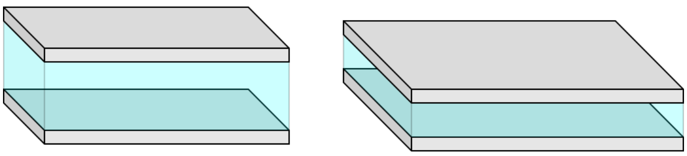


FIGURE 2.14: Two grey electrodes across a blue dielectric medium. Left: Uncompressed state. Right: Compressed state exhibiting larger electrode areas and a thinner dielectric thickness.

2.3.1.3 Magnetic

2.3.1.4 Optical

2.3.1.5 Acoustic

2.3.1.6 Pressure mapping technology comparison

2.4 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

2.4.1 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 [].

2.4.2 Modelling Conduction mechanism

The typical fabrication process stated in Section 2.4.1 for CPECs shows that the dispersion of conductive particles will always vary.

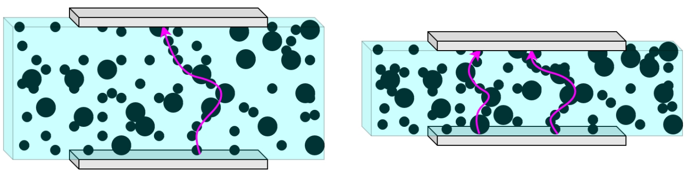


FIGURE 2.15: Two grey electrodes across a CPEC cuboid showing enlarged black conductive particles within a blue polymer matrix. Left: An uncompressed CPEC. Right: A compressed CPEC.

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 [117, 118]
   (b) Inherent particle conductivity
2. Conductive particle dispersion [119]
   (a) Inter-particle distance distribution
   (b) Particle agglomeration distribution [120]

   (c) Isotropy/anisotropy [121]
   (d) Sedimentation [122]

3. Elastomeric matrix
   (a) Viscosity
   (b) Elastic modulus
   (c) Dielectric permittivity

4. Impurities

5. Voids

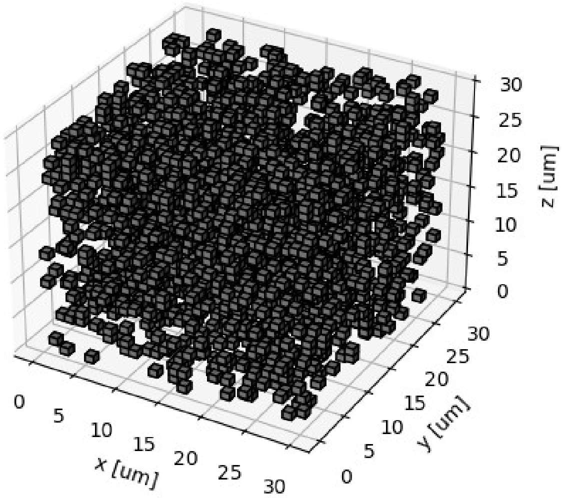


FIGURE 2.16: Example of a randomised cube percolation with a volume percentage of 8% of particles

Microscale models for CPECs and the relationship between particle and electric charge motion are often computationally heavy, overly idealised, and non-invertible [123]. A microscale model example can be seen in Figure 2.16. 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[124].

Electrical DC conduction through a CPEC occurs using two main mechanisms, Coulomb conduction and quantum tunneling [125–128]. 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 [129, 130].

Electrical AC conduction can occur through a CPEC through capacitive means depending of particle spacing[**?**].

The use of soft piezoresistive composites

Chapter 3

A Simple Conductive Elastomer Composite Material with Complex Behaviour

3.1 Introduction

As discussed in Section **??** conductive particle elastomer composites are desirable for soft sensor and actuator applications for a variety of reasons. However, it is crucial to understand the electromechanical behaviour of these composites if we wish to create complex control systems with such materials. Although conductive particle elastomer composites are a simple concept of dispersing particles throughout an elastomeric matrix, the electromechanical behaviour is not well understood on a macro or micro-scale. This section endeavours to understand the material behaviours of carbon black silicone rubber composites on a macro-scale to help create better inverse models so that the material can be used more accurately as a stress and/or strain sensor.

3.2 Material Imaging

3.3 Stress and Resistance Relaxation For Carbon Nanoparticle Silicone Rubber Composite Large-Strain Sensors

From a conference paper presented at IDETC-MESA 2021

Carbon nanoparticle-silicone elastomer composites are stretchable conductive materials with diverse applications such as, highly elastic strain sensors [108, 131, 132], dielectric elastomer actuators [133, 134] and electromyography electrodes[132, 135, 136]. Understanding the dynamic resistance relaxation characteristics of carbon black (CB) polydimethylsiloxane (PDMS) elastomer composites would improve performance in fields which require high efficiency of space, power and accuracy, such as the devices used in biomedical and aerospace fields. Unlike many common strain gauges, CB-PDMS composites can have strains of over 300% without yielding[137] depending on the type of PDMS and CB used and the method of fabrication.

Some characteristics of CB-PDMS composites which make it suitable for strain sensors include that, the material is relatively inexpensive and readily available; non-toxic and is bio-compatible; and has a significant and readily measurable resistance change when stretched. Whereas, alternatives to CB nanoparticles, such as carbon nanotubes[138, 139] and metallic particles[95, 140], have been seen to be more carcinogenic than the CB alternative[141–143]. The fabrication of the CB-PDMS composite requires a degree of optimisation to ensure that the carbon particles are adequately dispersed to ensure high conductivity and high yield strength of the material. More importantly the homogeneous dispersion of carbon black particles means better repeatability of experimental results and more accurate models for the eventual applications of CB-PDMS composites. A sufficiently comprehensive model of how the resistivity changes with strain has not yet been developed. A limitation of using this material as a strain sensor is the non-linearity of the material above a certain strain value, at which the composite's resistivity diverges towards a highly insulative value within the giga-ohms range. This non-linear behaviour of CB-PDMS can be used as a mechanically activated switching device[133]. If modelled, this non-linearity could extend the range of strains that can be measured.

While previous work from our research group [144, 145] has focused on the response to quasi-static and low speed behaviour, these materials show dynamic effects where resistance depends on the speed of stretching. The characterisation investigated for the CB-PDMS sensor involves understanding the relationship between the mechanical stress relaxation, electrical resistance relaxation and strain in time. A difference in time constants between the stress and resistance relaxations have been noted before in literature[138, 139, 146, 147], but never accurately modelled with the physical theory explained. The current limitations of predictability and repeatability of resistance relaxation hinders the accuracy of fitting models. An understanding of this resistance relaxation phenomena would mean an accurate model could be made to predict the relationship between stress, strain and resistance within a CB-PDMS composite. Finding this relationship model would also allow us to understand the limitations of using this composite in sensing applications and also the use of the material in dielectric elastomer actuators, whereby the material can be used simultaneously as an actuation excitation electrode and a strain sensor. The composite material can also be used in human motion measurement as a skin stretch sensor. Understanding these characteristics may give rise to new applications of the composites material, for example, if the resistive relaxation properties of the material were known, it could be used as a mechanically activated timing device. An oscillatory flexible dynamic circuit has been demonstrated when mimicking the motion of a caterpillar as shown by Henke et al.[133], where the resistance relaxation modelling is be useful for more accurate electrical circuit dynamics. The theory behind mechanical stress relaxation is widely known and has been modelled using a variety of mathematical models [19] depending on the material modelled. The research discussed will focus primarily on only tensile stress of the specimen, and how it relates to the electrical resistive relaxation.

BACKGROUND

The Composite

The CB-PDMS composite was composed of carbon black powder(Vulcan XC 72R, average particle size: 50 nm, typical bulk density: 96 kg/m^3) and two part Pt cured

PDMS(Smooth-On Dragon Skin 10 NV). This grade of PDMS was chosen due to the following characteristics [148]:

1. Low elastic modulus, E, of 186 kPa
2. Tensile strength, σ_y of 2.75 MPa
3. Low mixed viscosity, η, of 6,000 cps

The volume resistivity of pure carbon black powder itself is between 10^{-1} and 10^{2} Ωcm depending on how densely the particles are packed and the purity of the CB[108]. The ability of a carbon black matrix embedded within a highly insulative PDMS substrate to become conductive is determined mainly by the dispersion of the CB particles, and the tunneling that occurs between conductive CB and insulative PDMS bodies within the material volume[108, 139]. The composite being created must be highly conductive without compromising the elastic modulus and yield strength of the material. From percolation theory observed in literature [108] there is a threshold volume percentage of CB required to ensure that conductivity is maintained with certainty throughout the composite volume within the linear volume resistivity region. The percolation threshold for our composite was is difficult to predict analytically due to the unknown configurations of aggregates and agglomerations formed by the CB within the composite material. Experimentally we found that a CB volume percentage of 7.5% or greater meant the composite material had a resistivity of less than 3.5 kΩcm consistently with the fabrication method used.

The Mechanics

It is known that PDMS composites are viscoelastic materials and clearly exhibit the three traits of a viscolelastic material[19]: stress relaxation, strain creep and stress-strain hysteresis. Stress relaxation is effect observed when a step input of strain is applied to a material and there is a transient stress decay response which converges to a steady state value. A commonly used model for viscoelasticty is the generalized Maxwell body model of order n shown in Fig. 3.1.

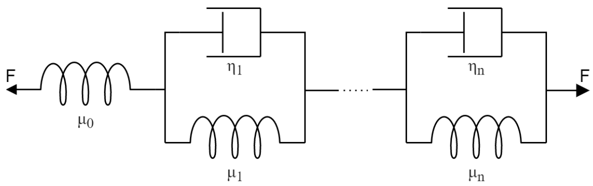


FIGURE 3.1: MECHANICAL SPRING DASHPOT DIAGRAM OF THE GENERALIZED MAXWELL BODY MODEL ADAPTED FROM FUNG ET AL.[19]

In Fig 3.1 F is the force applied to the material and μ and η values represent the spring and damping component constants, respectively. The stress relaxation function for this model is found in Eqn. 3.1, for, n, serial repeating units.

$$G(t) = a_0 + \sum_{i=1}^{n} a_i e^{-t/\tau_i} \tag{3.1}$$

Where a_0, a_i are the magnitudes of relaxation and τ_i are the relaxation decay time constant components. All of the constants a_0, a_i, and τ_i are functions of η and μ.

We initially assume that there is a relationship between the stress relaxation and resistance relaxation of the material. However the generalized model can easily over-fit the data, if n is too high, due to it's generality.

MATERIALS AND METHODS

Composite Fabrication

The first step in fabrication was to mix the CB nano-powder with the silicone part A (the liquid PDMS elastomer) using a Kurabo KK-50S planetary mixer. A mixing function was used with specific rotational velocities and times for each axis, which was well suited towards de-aeration and viscous particle mixing. The material was then mixed with the silicone part B (the liquid PDMS elastomer cross-linker) using the same planetary mixing function to ensure adequate dispersion of the CB particles throughout the PDMS volume as well as de-aeration.

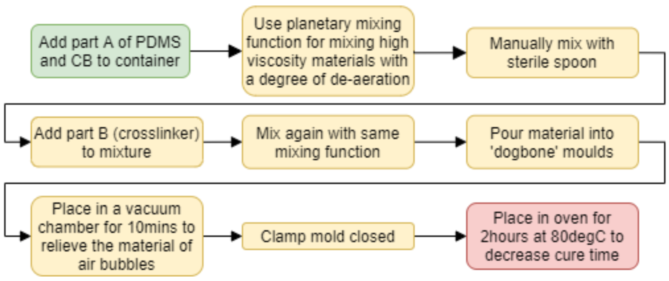


FIGURE 3.2: THE STEPS INVOLVED IN CREATING THE CB-PDMS COMPOSITE MATERIAL

For the fabrication of the CB-PDMS specimens, a standard dog-bone shaped mould was developed for the mixed CB-PDMS to cure in, based on ASTM standard D412[149]. Before the mould was clamped shut the composite filled mould was immediately placed in a vacuum chamber for ten minutes to de-aerate the still liquid, curing CD-PDMS mixture. The specimen was placed in a controlled oven at a temperature of 80 °C for a two hours to maintain the repeatability of the curing stage of the fabrication process. The temperature at which the silicone contributes towards the elastic modulus and yield strength of the material, with increasing curing temperatures giving increasing elastic moduli and decreasing yield strength values.

Measurement

A custom test measurement device was made for measuring the desired characteristics of the CB-PDMS material, so that parameters driving the data collection could be easily altered. The strain, stress and resistivity of the specimen were measured in in parallel. The setup included the use of a 500 gram loadcell (HT sensor - TAL221) in combination with a linear actuator stage driven by a NEMA23 stepper motor and an source measurement unit (Keithley 2634B SMU). A custom electrode clamp mechanism was designed to fix the electrodes onto the test specimen during the straining of the specimen. This consists of two copper plates sandwiching the composite material at each end of the dogbone test specimen.

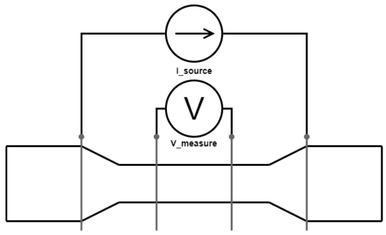


FIGURE 3.3: THE COMPOSITE DOGBONE TEST SPECIMEN PIERCED BY 4 PIN ELECTRODES. THE OUTER AND INNER ELECTRODES CONNECTED TO AN SMU CURRENT SOURCE AND VOLTMETER RESPECTIVELY

Two configurations of resistance measurement were tested, a two wire and a four wire method. The two wire measurement method used two electrodes which also clamped the test specimen at each end. It was observed that compressive strain applied to CB-PDMS composite will increase the resistivity of the specimen in a similar fashion to tensile stress. Only a compressive strain was applied to the material by the clamps such that the material would not slip during tensile testing and not deform giving erroneous resistance results. The Poisson's ratio of the material which was found experimentally to be 0.29 for both CB percentages. The two wire method used a controlled current source in parallel with a voltmeter attached to the same two electrodes to derive a resistance. The four wire method uses four pin electrodes as seen in Fig. 3.3. The four wire method applies a constant current source through the outer electrodes and uses a voltmeter on the inner two electrode to determine the resistance and hence resistivity of the material. The four wire electrode configuration meant that the resistivity had a smaller signal to noise ration compared to a two wire alternative.

Metallic pin electrodes were selected over copper clamp and conductive adhesive alternatives as they deformed the material the least, had a consistently low specimen-electrode contact resistance, and did not slip during test sequences. The inner pin electrodes were symmetric about the centre and placed 20 mm apart with the outer pin electrodes being 40mm apart as shown in Fig. 3.4.

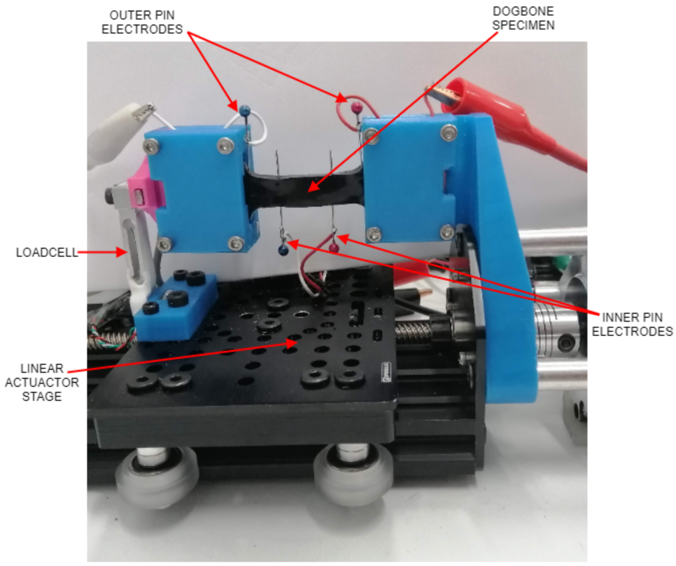


FIGURE 3.4: PHOTO OF TEST MEASUREMENT SETUP

The measurements were completed using finite pulse trains of strain to ensure repeatability of the models were consistent across varying experimental parameters. If this material is used as a sensor the model fitted to the stress relaxation must hold over many consecutive tensile strain events. As these materials are intended as large strain sensors, the strains tested in this work was 10%, 20%, and 30%. This strain percentage is higher than commonly used constantan strain gauges, which typically have a maximum strain of approximately ±3%[150], with traditional metal alloy based strain gauges often having significant plastic deformation after less than 10^4 cycles[150] at 3% strain.

RESULTS AND DISCUSSION

Viscoelasticity

All of the specimens fabricated indicated a degree of viscoelasticity shown by the hysteresis seen when loading and unloading the material with 30% tensile strain in Fig. 3.5. The 0, 7.5, and 10 w.t.% CB specimens have average elastic moduli, as measured in the loading phase, of 205.2 kPa[1], 321.4 kPa, and 342.1 kPa, respectively. The hysteresis loop seen in the 10 w.t.% CB sample has a larger hysteresis loop showing that there is increased viscous/damping compared with the other two specimens percentages of CB. The pure PDMS specimen had no discernible hysteresis from the data as shown in Fig. 3.5. The difference in hysteresis and hence viscolelastic properties, across the specimens will lead to different stress relaxation properties across the three composite materials.

[1]Different from the 186.2 kPa elastic modulus specified by the manufacturer due to the temperature accelerated curing method used

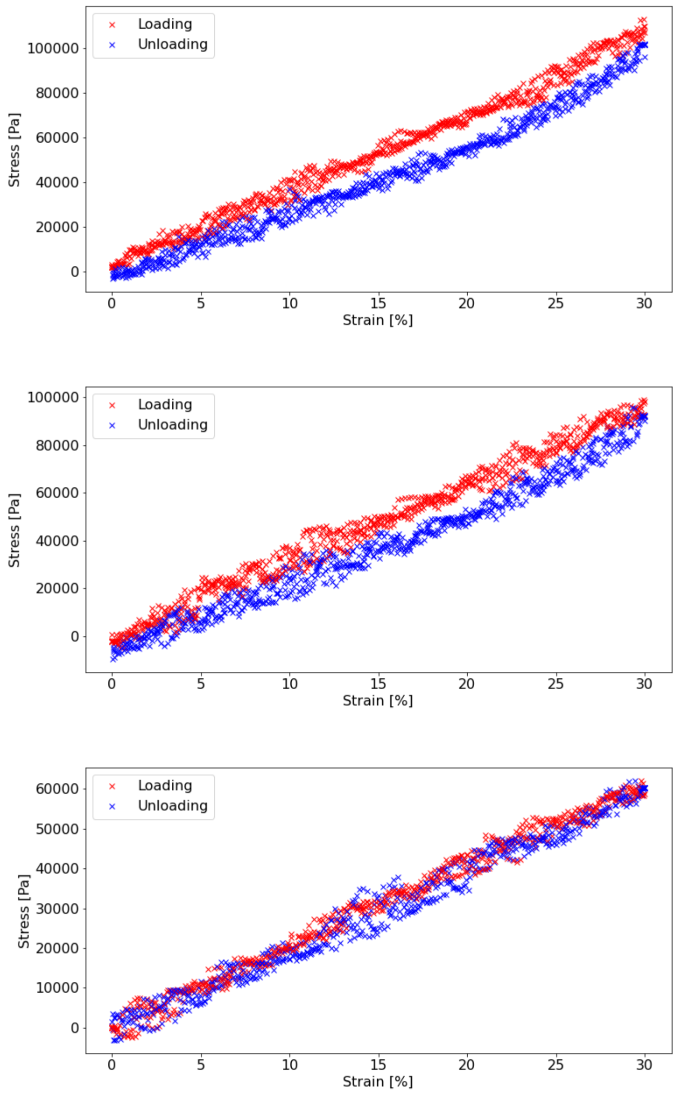


FIGURE 3.5: THE LOADING AND UNLOADING OF 30% STRAIN ON A COMPOSITE TEST SPECIMENS WITH CB WEIGHT PERCENTAGES FROM TOP TO BOTTOM OF 10%, 7.5%, AND 0% WITH DATA COLLECTED OVER FIVE LOADING AND UNLOADING CYCLES

Resistance Relaxation Model Fitting

The initial model chosen to fit the stress and resistance relaxation data was the generalized Maxwell body model shown in Fig. 3.1 with n = 3 cascading elements using Eqn. 3.2 to fit the model. Fitting the data using Levenberg–Marquardt non-linear least square algorithm over 30 data sets showed an instability with the algorithm using this model. When feeding the previously fitted stress relaxation model constants as initial conditions for the fitting of the next stress relaxation data set, the values of the constants diverged exhibiting signs of overfitting. This divergence of the model constants meant that they had a large standard deviation showing the model was changing significantly each iteration of fitting. Hence a more simple model using Eqn. 3.1 with n = 2 was used to fit the stress relaxation data to Eqn. 3.3 with lower standard deviation of the model constants. Conversely when the resistance relaxation model analogous to stress relaxation model, shown in Eqn. 3.4, was fitted to the resistance relaxation data there was a stable fit with a better goodness of fit.

The decay time constants of the two models are different with the resistance having an longer overall decay which can clearly be seen in Fig. 3.6. Below in stress relaxation models $G_{1,2}(t)$, shown in Eqn. 3.2 and 3.3, the constants a_{0-3} and τ_{S1-S3} represent the components of magnitude and time decay of the stress relaxation, respectively.

$$G_1(t) = a_0 + a_1 e^{-t/\tau_{S1}} + a_2 e^{-t/\tau_{S2}} + a_3 e^{-t/\tau_{S3}} \tag{3.2}$$

$$G_2(t) = a_0 + a_1 e^{-t/\tau_{S1}} + a_2 e^{-t/\tau_{S2}} \tag{3.3}$$

Analogously for the resistance relaxation function $H(t)$, the constants b_{0-3} and τ_{R1-R3} represent the components of magnitude and time decay of the resistance relaxation, respectively.

$$H(t) = b_0 + b_1 e^{-t/\tau_{R1}} + b_2 e^{-t/\tau_{R2}} + b_3 e^{t/\tau_{R3}} \tag{3.4}$$

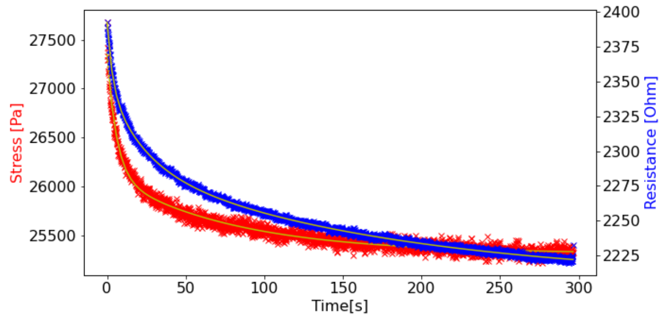


FIGURE 3.6: COMPARING THE RELAXATION DECAY TIME CONSTANTS OF STRESS AND RESISTANCE FOR A 7.5 W.T.% CB-PDMS COMPOSITE AFTER A 10% STRAIN STEP INPUT AND FITTING GENERALIZED MAXWELL BODY MODELS TO EACH.

The mean magnitude and decay time constants for the resistance and stress relaxations using 30 relaxation periods to fit the models to are given in table 3.3. The data gathered show that the stress relaxation time constant values decrease with an increasing carbon black percentage, indicating that all constants in Equations 3.4 and 3.3 are also functions of the carbon black percentage.

TABLE 3.1: FITTED CONSTANTS AND THEIR MEAN, μ, STANDARD DEVIATION, σ, AND COEFFICIENT OF VARIATION, CV, VALUES FOR 0%, 7.5%, AND 10% CB-PDMS COMPOSITE SPECIMENS USING EQUATION 3.3.

Stress Model

Constant	μ	σ	CV
0 % CB Specimen			
a_0	20344.71	42.61	0.20%
a_1	387.28	59.86	15.45%
a_2	526.82	57.65	10.94%
τ_{S1}	72.08	23.46	32.54%
τ_{S2}	5.77	1.48	25.75%
7.5 w.t.% CB Specimen			
a_0	25363.89	33.62	0.13%
a_1	802.32	43.59	5.43%
a_2	1242.32	52.67	4.24%
τ_{S1}	71.01	9.49	13.37%
τ_{S2}	5.79	0.65	11.32%
10 w.t.% CB Specimen			
a_0	32303.01	165.62	0.51%
a_1	1071.38	54.32	5.07%
a_2	1649.82	47.31	2.86%
τ_{S1}	84.07	10.55	12.54%
τ_{S2}	6.52	0.74	11.35%

TABLE 3.2: FITTED CONSTANTS AND THEIR MEAN, μ, STANDARD DEVIATION, σ, AND COEFFICIENT OF VARIATION, CV, VALUES FOR 0%, 7.5%, AND 10% CB-PDMS COMPOSITE SPECIMENS USING EQUATION 3.4.

Resistance Model

Constant	μ	σ	CV
7.5 w.t.% CB Specimen			
b_0	2154.31	52.68	2.44%
b_1	81.13	5.39	6.65%
b_2	56.37	3.67	6.52%
b_3	42.16	3.42	8.12%
τ_{R1}	181.10	33.57	18.54%
τ_{R2}	22.84	3.81	16.71%
τ_{R3}	3.46	0.56	16.35%
10 w.t.% CB Specimen			
b_0	1649.55	97.44	5.90%
b_1	55.19	8.85	16.04%
b_2	77.39	12.23	15.80%
b_3	38.35	9.47	24.69%
τ_{R1}	169.63	61.72	36.38%
τ_{R2}	21.85	9.66	44.21%
τ_{R3}	3.02	1.59	52.72%

Our aim was to prove the hypothesis that the stress relaxation time constant is different to that of the observed resistance relaxation and able to be modelled mathematically. The apparent difference in time constants and the fitting of the data to two different equations show that the stress relaxation is not linearly related to the resistance relaxation shown clearly in Fig. 3.6. To display the non-linear relationship between the stress and calculated resistance within the material they are plotted against each other over 30 sequential relaxation periods of 300s. The non-linear relationship between stress and resistance changes over time for each relaxation as shown in Fig. 3.7, where the data for the first relaxation is displayed in green and the last relaxation in blue.

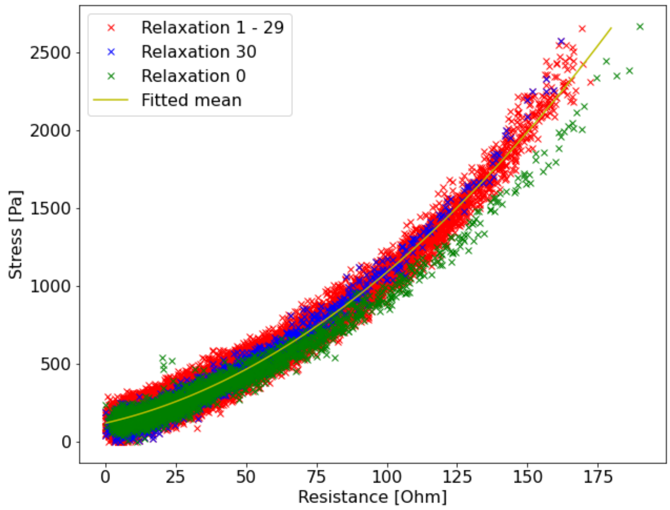


FIGURE 3.7: COMPARING RESISTANCE AND STRESS RELAXATION DATA AGAINST EACH OTHER OCCURRING DURING 30 PULSES OF A 10% STRAIN STEP INPUT FOR A 7.5 W.T.% CB-PDMS COMPOSITE

The stress-resistance relaxation data was fitted to a generic second order polynomial of the form,

$$\sigma(R) = aR^2 + bR + c \tag{3.5}$$

where σ is stress, R is the calculated resistance. When fit to the latter 15 cycles of a 30 cycle 10% strain pulse train of stress relaxation data we get the constant values for a, b and c.

TABLE 3.3: FITTED CONSTANTS AND THEIR MEAN, μ, STANDARD DEVIATION, σ, AND COEFFICIENT OF VARIATION, CV, VALUES FOR 7.5%, AND 10% CB-PDMS COMPOSITE SPECIMENS USING EQUATION 3.5

7.5 w.t.% CB Specimen			
Constant	μ	σ	CV
a	0.055	0.006	11.1%
b	4.146	1.058	25.5%
c	121.845	16.338	13.41%
10 w.t.% CB Specimen			
Constant	μ	σ	CV
a	0.098	0.007	7.48%
b	6.374	0.757	11.87%
c	155.812	38.753	24.87%

Strain Velocity Resistance Relationship

A narrow peak in the apparent resistance has been observed in the collected data when changing from 10% strain to a zero strain. This peak is not present in the stress plot, hence is a proposed characteristic of electrical behaviour only as a function of strain. In previous literature, the effects of the rate of change of strain on apparent resistance of the CB-PDMS material has not been modelled or shown. When the material has finished a tensile cycle of strain and is returning a zero strain state the a component of the resistance, R_p, can be modelled with a second order polynomial. When differentiated, this peak gives a linear function in a similar form of the linear strain curve seen in Fig. 3.8. Hence we form an equation which relates a component of resistance,

$$\frac{dR_p}{dt} = E(\varepsilon)t + c \tag{3.6}$$

where E is a function of strain, $\varepsilon(t)$, and c is an offset bias determined by the initial strain condition. To show the strain velocity resistance relationship, more strain pulse

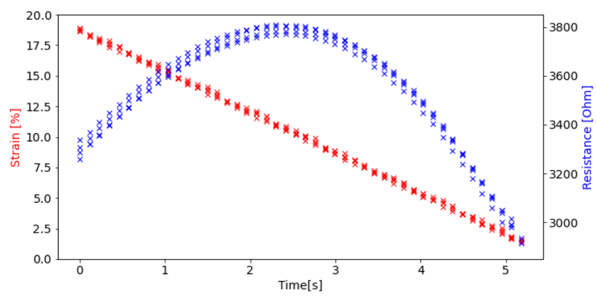


FIGURE 3.8: STRAIN VELOCITY RESISTANCE RELATIONSHIP SHOWING THE SPECIMEN IS RETURNING TO A 0% TENSILE STRAIN STATE FROM 10% AT A STRAIN RATE OF 80mm/s FOR FOUR TESTS FOR A 7.5% CB-PDMS SPECIMEN

train tests of 20% strain were completed. Using 20% strain allowed us to see a sufficient number of data points to observe a trend. The pulses had four repetitions with a range of strain velocities, $\dot{\varepsilon}(t)$, of 40, 80, 120 and 160 mms^{-1}. Using a 7.5 w.t.% CB-PDMS specimen we obtain a relationship that agrees with the strain resistance component equation 3.6. As $\dot{\varepsilon}(t)$ increases through strain speeds so does the magnitude of the resistance peak (i.e. maximum height of the resistance peak - the previous steady state of value resistance) of 400, 510, 569, and 641 Ω for $\dot{\varepsilon}(t)$ of 40, 80, 120 and 160 mms^{-1} respectively. A new model is required which can accurately reproduce the additional decay time constant and small peak features seen in the resistance relaxation data, so that the resistance can inversely calculate the strain in the material.

Repeatability

The resistance relaxation model must be predictable over many strain cycles for use within many high stretch strain sensor application. If the resistance relaxation changes over time this needs is to be modelled. Each test sequence showed that there was a downward trend in the calculated magnitude of resistance for each pulse over time. This downward trend is hypothesized to be due to the accumulation of charge within material over time generated by current source, and was mitigated by using an alternating polarity measurement technique. The reversible current source helped to mitigate the capacitive effects seen, but a general downward trend in resistance was still observed as shown in Fig. 3.9. For every sufficiently long test sequence the material reaches a steady state, after a finite amount of time. The capacitance read across the inner pin electrodes of the material decreased with increasing strain as shown in Table 3.4.

TABLE 3.4: AVERAGE INNER ELECTRODE CAPACITANCES, C_i, MEASURED FOR VARIOUS STRAIN, ε, VALUES USING A 7.5 W.T.% CB-PDMS COMPOSITE, MEASURED USING AN LCR METER AT 1kHz AND 10kHz

$\varepsilon[\%]$	0	10	20	30
$C_i[pF]$	53	32	24	20

The generalized Maxwell model has been applied to predict the stress relaxation of the CB-PDMS composite and analogously the resistive relaxation seen, which successfully explains a significant fraction of the resistance relaxation seen for a positive strain step input. However, a sudden peak of resistance when changing from +10% strain to 0% is not yet explained, and consideration of temperature and strain history[19] will be useful to confirm the simple mathematical model given as Eqn. 3.6.

In this work, mixing has been performed using a planetary mixer. It has been shown in other works [108, 151] that other mixing methods, such as using a sonication bath and the addition evaporateable solvents, can yield better particle dispersion. A higher degree of CB particle dispersion has also been shown to alter the viscoelastic creep properties [151], and is therefore likely to affect the time constant of resistance.

CONCLUSIONS

In order to improve the accuracy of dynamic strain measurements with CB-PDMS composites a stress and analogous resistance relaxation model was formed. The generalized

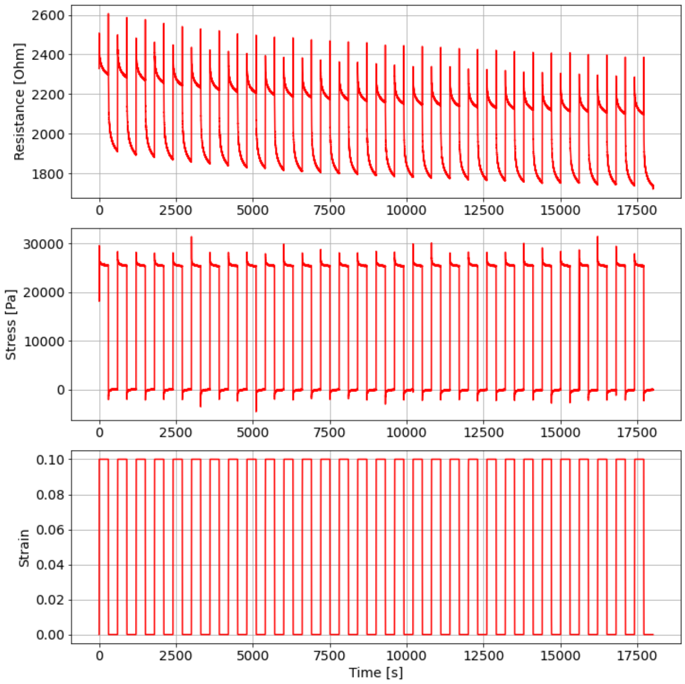


Figure 3.9: A typical test sequence of a 30 pulse strain train recording the calculated resistance and stress of a 7.5 w.t.% CB composite

Maxwell model, Eqn. 3.3 was used to fit to the stress relaxation data for three specimen with CB weight percentages of 0, 7.5% and 10%. The CV of the stress relaxation magnitude constants $a_0 - a_2$ were found to be consistently smaller than the CV of the stress relaxation decay time constants τ_{S1} and τ_{S2}, with maximum CV values of 15.45% and 32.54% respectively. All of the stress relaxation model constants increased with increasing weight percentage of CB.

After modelling the stress relaxation, an analogous resistance relaxation model, Eqn. 3.4 was formed and fitted to, displaying similar attributes to the stress relaxation model fit with all of the model constants increasing with increased w.t.% CB. The CV of the analogous resistance relaxation magnitude constants $b_0 - b_3$ were found to be consistently smaller than the CV of the stress relaxation decay time constants $\tau_{R1} - \tau_{R3}$, with maximum CV values of 16.04% and 44.21% respectively.

A model relating the resistance and stress relaxation has been developed using a second order polynomial with all of the constants a, b, and c increasing with increased weight percentage of carbon black. With the models developed we have shown that the apparent resistance relaxation can be modelled, which will enable more accurate estimation of dynamic strain when these materials are applied as sensors.

3.4 A Piece-wise Approach to Modelling Carbon Black Silicone Rubber Composites

One method for understanding the transient behaviour of CPECs is to create a classification system and determine mathematical relationships that can be matched to these transient event. Mersch et al. have classified several shoulder events and the related deformation events, compressive, tensile, and bending. These transient peaks have been observed by several researchers using the similar CBSR materials, however there is no conclusive mathematical model relating these transient peaks to strain in time. This section aims to further classify these transient events and provide a mathematical relationship, for future use with model fitting methods.

3.4.1 Rising Edge Step Response

3.4.2 Falling Edge Step Response

As shown in Section 3.3 there has been a mathematical relationship observed between the falling edge of a strain input and the resultant resistance peak. Consequently a parameter fit study has been completed to determine how to predictably control the resistance peak through a controlled strain input. We can see a repeated property in Figure 3.8 whereby the derivative of the resistance signal seems to be equal to the strain curve.

To prove that there does exist a mathematical relationship between the two signals the relationship first each signal is given a generalised formula. The resistance signal is parabolic Equation 3.7.

$$R_p = A(t - H)^2 + K \tag{3.7}$$

Where strain rate changes the vertical shift, K, time shift, H, and concavity, A, of the parabola.

3.4.3 Strain Rate

3.4.4 Saw Tooth Response

3.5 Characterising Hysteresis

Chapter 4

A Novel Mixing Method for 3D Printing Conductive Particle Elastomer Composites

4.1 Introduction

Chapter 5

An Improved an Electrical Impedance Tomography Based Artificial Soft Skin Pressure Sensor

The content from this chapter is predominantly from the manuscript published in the journal Sensors and Actuators A Physical - E-Skin special issue.

ABSTRACT

Using electrical impedance tomography (EIT) to drive a pressure mapping device shows great potential, due to the customisability of the sensing domain and the non-invasive nature of the boundary electrodes. A pressure mapping system has been developed in this work that uses a silicone carbon black nanoparticle sensing domain, giving the domain with a comparable softness to human skin tissue. To take this technology into a commercial application the performance of such an EIT-based sensor must be quantifiable and repeatable. In this work a series of experiments were repeated for various load locations, strains, and carbon black percentages. Capturing this data gave insight into the how the sensing domain performs over time and captured the transient events limiting the sensor. Metrics were determined to quantify the sensor's spatial resolution. A quasi-static conductance-force model of the material was developed with an accuracy of $\pm$0.78 N. One important metric is temporal resolution, as it is the least quantified performance metric in literature, however can be the most important for some applications. For the sensor domains tested, average settling times of between 19.0 - 44.5 s and 22.5 - 36.0 s were determined for 8 and 9 wt% CBSR samples. A series of randomised test loads gave similar spatial performance results to the structured experiments. This sensor platform shows promise for future applications, with further materials development and processing of data the rise of an artificial biomimmetic pressure sensitive skin is imminent.

5.1 Introduction

Approximately 1 billion years after the first animals developed mechanosensation [152], evolution has allowed humans to detect pressure through the use of many mechanoreceptors lying within the skin and other organs. Two mechanoreceptors which are desirable to emulate human touch are Merkel's disks and Meissner's corpuscles [153]. Both of which are ubiquitous in human hands and lips for high spatial resolution, low pressure and low frequency touch/pressure events [154]. These mechanoreceptors in a human hand enable object identification and closed loop fine motor control.

With the creation of pressure mapping technology which has the similar soft mechanical properties and sensing qualities to that of human skin many commercial applications requiring human-like touch could be directly fulfilled. This work presents characterisation of a soft mapping pressure sensor which utilises electrical impedance tomography to map resistance changes and subsequently stress changes throughout a soft material surface.

The number of applications that require 2D pressure sensing using a soft surface is extensive. Such applications include: robotic gripper object detection, medical mattresses and cushions, limb prostheses and wearable robotics, sport equipment, smart furniture, and rehabilitation devices. The following characteristics are desirable for each of these applications: force sensitivity, low toxicity, cost-effectiveness, repeatability, and high elasticity. In this work, a system showcasing each of these desirable characteristics has been developed.

The sensor platform utilises a piezoresistive nanoparticle elastomer composite (PNEC) in a thin sheet topology to create an artificial sensitive skin. This artificial skin is composed of a highly elastic piezo-resistive material, and its deformation can be identified through electrical impedance tomography (EIT) for the reconstruction of the material resistivity image. Using 16 boundary electrodes, EIT facilitates the mapping of applied forces on this monolithic homogeneous material. Subsequently, an inverse model is applied to estimate compressive force loads on the material.

Understanding the electro-mechanical properties of the PNEC material is essential for creating an accurate dynamic sensor. When elastomeric composites with conductive particles, such as the PNEC, exhibit viscoelasticity, the degree of hysteresis varies based on the constituents of the composite material [155]. This viscoelasticity is a major limiting factor when using PNECs for EIT-based pressure sensing due to the frequency response lag introduced by the large transient effects seen in the material. In this work these transient phenomena are captured and characterised in the 1D and 2D compressive stress cases.

Various methods and topologies of 2D pressure mapping sensors can be employed for a 2D resistivity measurement. However, many of these methods involve intrusive and intricate electrode placement within the material domain [156–161]. Since the materials utilised in this study are soft, the utilisation of relatively rigid metal electrodes distributed throughout the material would significantly alter the material's electromechanical deformation response. This necessitates use of the non-invasive method imaging method EIT.

5.1.1 EIT Background

To estimate the 2D resistivity of the PNEC a technique called electrical impedance tomography (EIT) was used. EIT allows the generation of a map of impedance values of a thin cross section of a domain under test (DUT). EIT uses a set of boundary electrodes to pass known electrical currents and measure voltages along the boundary of the DUT. From these known current injections and voltage measurements, an ill-posed inverse problem can be defined. To obtain an EIT image reconstruction three key steps are required: data acquisition, forward modelling, and inverse problem solving. A constant current can be employed to capture solely the resistance values of the DUT or as an AC signal to sweep through a range of frequencies to capture impedance data.

The forward problem in EIT is a well-posed mathematical problem, so linear algebra can be employed for obtaining electric field data for a DUT of known conductivity and a known current injection. Utilising a mesh-based coordinate system and FEM, proves to be an efficient solution for the forward model, accommodating diverse shapes. Solving the forward problem entails applying Maxwell's electromagnetic formulae to determine how an electric field would propagate through the DUT, considering the DUT conductivity. An initial estimate of the DUT resistivity is necessary for the first step of an EIT algorithm. Once the EIT forward model is solved, EIT inverse problem can be solved iteratively using the forward problem's solution. This inherently unstable problem requires optimisation algorithms and regularisation to create and linearise a solution [162–165].

Once the EIT reconstruction algorithm has been tuned for an application as desired, often post-processing is completed on the reconstructed EIT image for filtering and to capture data specific for the application. A consortium of experts in the field of medicine and biomedical imaging have constructed metrics for quantifying the quality of an EIT reconstruction as shown by Adler et al. [166] and their GREIT (Graz consensus Reconstruction algorithm for EIT) performance metrics. Researchers who have used EIT pressure sensing purposes have also developed performance metrics, most of which agree with the GREIT metrics [106, 158, 167**?** –170]

5.1.2 Related Work

Artificial skins are not a novel subject there are many different methods for localising loads in two-dimensions on a soft domain. The limiting factors found with non-EIT based methods of pressure mapping were the size discretely sensed regions, also known as sensels, is limited by various factors in the fabrication process and the bulk of electrode wires required. This bulk is exemplified high electrode-to-sensel ratio. Example load mapping technology include, optical [171–173], piezoresistive [160, 174, 175], capacitive [176], and magnetic [177]. Each of which have been compared in Table **??**.

Other attempts at creating artificial sensitive skins using EIT have been shown in a review by Silvera-Tawil et al. [158]. This review provides adequate evidence to display interest in the field; however, there is still no commercial EIT-based pressure sensor that is comparable in terms of spatial and temporal resolution, to commercially available non-EIT-based 2D pressure sensors. One of the earliest applications of EIT to an elastic piezo-resistive domain was achieved by Knight and Lipczynski [178] in 1990. Since this application, several other similar systems have been created using EIT and similar pressure sensitive fabrics or elastomeric materials[106, 158, 170, 179–182]. A comparison of similar devices is given in Table ??. None of these researched devices focus on using a material with similar softness, and quantify the stress data captured in real-time like this work. All of the referenced 'EIT' soft sensors employ electrical resistivity tomography (ERT), however, the term ERT is most commonly associated with geological subsurface imaging applications, henceforth, EIT is be used in place of ERT in this work.

5.2 Methodology

To substantiate the applicability of Electrical Impedance Tomography (EIT) with a monolithic PNEC sample, we fabricated the material for testing. The material needed to adhere to specific requirements: highly elastic, high yield strength, low resistivity, high piezoresistivity, non-toxic, and be a low Shore hardness of 5A - 25A akin to human soft tissue [41, 158, 183, 184]. Additionally, a system of devices was devised to facilitate EIT measurements which concurrently captured force, strain, and timestamps for each measurement. Lastly, to evaluate the sensor's suitability for diverse applications, spatial, temporal, and localised force sensing performance metrics were quantified.

5.2.1 Fabrication

The fabrication of the piezoresistive composite materials, as described and justified in our previous work [155], involved dispersing 8 and 9 wt% of carbon black (CB) nanoparticles in a silicone rubber (SR) matrix. Because of the difference in fabrication processes seen in literature [185, 186] and degree of dispersion generating variability in the percolation, an iterative trial and error approach using the starting point found in literature was used to get 8 wt % and 9 wt % values for CB in SR. Within this range the material was sufficiently conductive while maintaining mechanical strength through sufficient elastomeric cross-linking. Previous research indicates that there is a weight percentage at which the gauge-factor/piezoresistivity is at a maximum within a similar range used in this work [187, 188]. The composite, designated as the domain under test (DUT), was created using 50 nm average diameter XC 72R CB nanoparticles (Cabot, Alpharetta, USA) in a Dragon Skin 10 NV silicone rubber matrix (SmoothOn, Macungie, USA). Homogeneous dispersion was ensured using an ARV-310 vacuum planetary mixer (Thinky, Tokyo, Japan).

TABLE 5.1: DUT mechanical characteristics and electrical characteristics

Sample	**CB wt** [%]	$\mathbf{R_{int}}$ [$k\Omega$]	**E** [kPa]
SR	0	$> 1 \times 10^9$	186.16
CBSR	8	18.1 ± 9.8	132.5
CBSR	9	4.5 ± 1.4	98.1

A CBSR sample showing the circular sensitive region with the pin electrodes developed is shown in Figure 5.1.

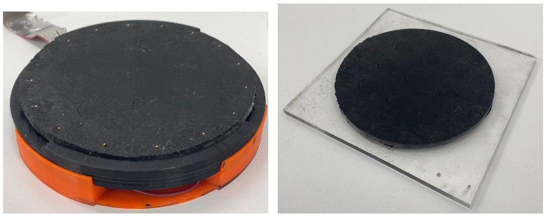


FIGURE 5.1: Left: Example of a CBSR sensing domain with gold pin electrodes penetrating material surface around the boundary on top of the rigid sensor holder (orange/black). Right: CBSR sensing domain.

5.2.1.1 Localised Stress Testing

Quantitative results are required for spatial quantification of the EIT image reconstructions. A cylindrical force applicator head with a diameter of 13 mm and area of 133 mm^2 was used to apply the nine compressive loads shown in Figure 6.7.

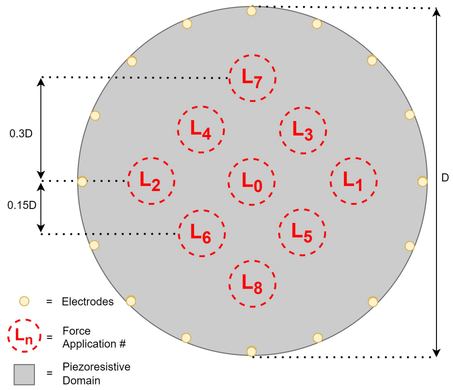


FIGURE 5.2: Load application areas used for compressive stress testing shown numerically in order of application.

5.2.1.2 EIT Measurement

At its core EIT usually requires a current or voltage source, one or multiple voltmeters, and a switching device. When integrating a mechanical pressure validation system a force applicator (CFA) and is also required to capture data simultaneously. The system architecture and DUT electrical connections are shown in Figure 5.4.

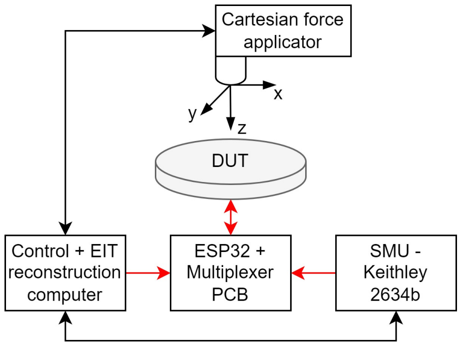


FIGURE 5.3: Architecture of the Cartesian force applicator setup with red arrows being analogue power lines and black arrows being digital data lines

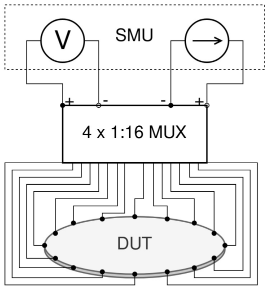


FIGURE 5.4: Wiring diagram for sensor connection to 4:16 multiplexer and SMU

5.2.2 1D Material Characterisation

Prior to utilising CBSR materials as a 2D pressure sensor, the piezoresistive properties were analysed in one dimension to establish resistance-stress/strain relationships for each CBSR sample. This 1D material testing was conducted using the Cartesian force applicator in conjunction with the SMU. The stress-strain relationship of the material was determined in previous work and shown in Table 5.1. The 1D analysis gave quantitative insight into the material resistivity response to strain in different known areas of the Device Under Test (DUT).

5.2.2.1 Quasi-static Piezoresistivity

To determine the piezoresistivity or gauge factor of the material, pin electrodes were pierced through the CBSR samples so that the pin electrodes were parallel and at a distance of 35 mm from each other. The pins were 2.5 mm from each end of the sample. A 2634b source measure unit (Keithley, Solon, USA) was used to apply a constant current of 1 mA between the two pin electrodes while ten compressive loading cycles were applied. The ten loading cycles were applied at strains of 5, 10, 15, 20, 25, and 30%, with a duty cycle of 50% and period of 120 s. Loads were applied using the Cartesian force applicator with a 20 mm x 20 mm square flat force applicator head. The strain rate was kept at a constant 16.67%.s^{-1} to dampen the amplitude of transient effects as proven in previous work [155].

When using this material to estimate stress based on resistance, any transient effects that are not correlated between resistance and stress must be accounted for. To obtain a stress reading from this PNEC CBSR material using the quasi-static model given in Equations 5.1 and 5.2, transient events were ignored, so only the steady-state conductance of the material was utilised.

$$\frac{\Delta\rho}{\rho_0} = \alpha_\sigma \sigma + \beta_\sigma \tag{5.1}$$

$$\frac{\Delta\rho}{\rho_0} = \alpha_\varepsilon \varepsilon + \beta_\varepsilon \tag{5.2}$$

Where α and β are the linear fit parameters, σ is the compressive stress, ε is the strain, $\Delta\rho$ is the change in conductance, and ρ_0 is the original material conductance.

5.2.2.2 Transient Piezoresistivity

There are two main piezoresistive events that occur during these compressive stress pulse response experiments. They are the compressive loading and unloading transients. Both of which result in stress relaxation and resistance relaxation behaviour until a steady-state resistance is reached. The stress relaxation can be approximated by a generalised Maxwell linear viscoelastic model [189]. A two component model was found to fit all curves without overfitting, the relaxation model from a step input is given in Equation 5.3.

$$G_2(t) = a_o + a_1 e^{-t/\tau_1} + a_2 e^{-t/\tau_2} \tag{5.3}$$

Where $G_2(t)$ is the stress relaxation function, a_0 is the relaxation offset, a_1 & a_2 are the magnitude weightings for each time constant τ_1 & τ_2. Equation 5.3 was used analogously for the resistance relaxation. To ensure repeatability of the experiment the ten loading and unloading events were fitted using equation 5.3 to the each relaxation, then the R^2 goodness of fit was compared for all of the relaxations.

An important temporal characteristic of this system is the settling time, t_s, given a strain step input. In this system the resistance step response settling time was the time taken to reach and stay within a specified tolerance about final steady-state resistance, given a strain step input. The tolerance chosen is about the steady state was $\pm$ 15%.

5.2.3 Sensor Performance Metrics

To ensure that the resistance/conductance image reconstructions which will form stress maps are valid solutions, the quality of the reconstructions needs to be quantified computationally. The purpose of this section is to describe various metrics used for sensor validation. These metrics measure the spatial performance, temporal performance, and localised force sensing performance. Many of the spatial and temporal performance metrics have been taken and adapted from Adler et al. [166] and their GREIT (Graz consensus Reconstruction algorithm for EIT) performance metrics.

5.2.3.1 Pre-processing

To ascertain the occurrence of a piezoresistive event in the material, it is necessary to identify a change in resistivity that surpasses the noise floor level. This precaution is taken to differentiate between a stress compression event and potential noise artefacts. A threshold was established to eliminate the noise floor, thereby isolating the loading signal and any noise or artefacts generated by the loading signal(s).

A second threshold filter was implemented to compensate for the regularisation of the reconstruction using a percentage of the largest peak observed in the sensor image domain. The percentage threshold value used in previous work has been 25, 50, 60, 70, 75% of the maximum domain amplitude [158, 166, 167]. In this work percentage threshold masking have been applied for comparison. To validate the best percentage threshold, these thresholds were completed for a CBSR 8 and 9 wt% PNEC under nine successive loading events comparing the mean of the three main performance metrics given in Section 5.2.3.2. After these threshold masks have been applied to the 2D EIT images, blob(s) are observed as the sensed regions-of-interest. In this work the term 'blob' refers to an amorphous 2D shape made of several aggregated finite mesh elements. These blobs are usually observed after percentage threshold masking of an EIT image reconstruction.

5.2.3.2 Spatial Performance

The three main metrics of spatial performance are the centroid or centre of 'mass' error, E_{CoM}, the detected area overlap A_{OL} value, and the fit of the detected blob relative to the force input, the shape distortion, SD.

The E_{CoM} was found using:

$$E_{CoM} = \sum_{i}^{N_b} e_{CoM_i} \times \frac{e_i}{e_{total}} \tag{5.4}$$

Where N_b is the number of elements in the threshold masked blob, e_{CoM_i} is each individual element centroid, e_i is the i^{th} blob element resistance value and e_{total} is the sum of all of the blob element resistance values. This equation can be easily be inverted for images containing conductance elements in place of resistance elements. The nearer the E_{CoM} value is to zero, the better the reconstruction in regards to localising the sensed region.

The A_{OL} was found using:

$$A_{OL} = 100 \times \left[\frac{\left(\sum_i^{N_b} A_{e_i}\right)}{A_{FA}} + \frac{\left(\sum_i^{N_b} A_{e_i}\right)}{A_b} \right] \Big/ 2 : e_{CoM_i} \in \Omega_{FA} \tag{5.5}$$

Where A_{e_i} is the area of an element and Ω_{FA} is the domain of the force applicator area. A_{FA} and A_b are the areas of the force applicator and sensed region respectively. The closer the A_{OL} is to 100%, the better the overlap of the estimated and actual load application.

The SD was found using:

$$SD = \left[\sum_{j}^{N_p} \left(\|F_{CoM} - P_j\|_2 - r_{FA} \right)^2 \right] \Big/ N_p : P_j \in L_b \tag{5.6}$$

Where N_p is the number of mesh nodes on the perimeter of the blob, F_{CoM} is the force applicator CoM coordinates, P_j is a node in the set of blob perimetral nodes, L_b, and r_{FA} is the radius of the force applicator head. The *SD* is essentially the mean square error of the force applicator perimeter and sensed region perimeter taken radially from the force applicator centroid.

5.2.3.3 Temporal Performance

A core problem with using soft PNECs is the temporal resolution they can provide can be limited by the viscoelasticity in the material and how it interacts with the conductive network in the material. This section provides insight into how to determine the frequency response of the material, by observing the relaxation settling time of the material. The maximum time taken to reach a steady-state resistance after a stress event dictates the frequency at which the material can sense stress events. To determine the time taken to reach a steady-state, a variety of compressive stresses and strains at the loading locations, L_0-L_8, of each DUT are observed.

To determine whether the resistance relaxation observed in the 1D case matches the relaxation of the reconstruction in 2D, the 1D and 2D relaxation settling times were compared to validate whether similar frequency responses were being observed. The EIT measurement equipment was designed such that the reconstruction frequency of 0.4 Hz can capture these transient events observed triggered in the CBSR samples, which are typically in the order of tens of seconds.

From the blob localisation described in section 5.2.3.2 the resistance relaxation data was extracted by plotting the sum of the blob resistance. Similar to the 1D case, the relaxations for stress loading, resistance loading, and resistance unloading scenarios are captured and compared for each CBSR with CB weight percentages of 8 and 9 wt%. The blob resistance relaxation was compared to relaxation deduced from the total image domain resistance to ensure that the only the transient event from the area being loaded was being observed.

5.2.3.4 Localised Force Sensing Performance

As discussed in Section 5.2.2.1, a quasi-static function, Equation 5.1, has been generated that gives a stress based on 1D steady-state conductance measurements. This quasi-static function was applied to the DUT 2D reconstruction to obtain a stress map of the material. To determine the minimum detectable stress of the sensor the conductance noise floor values were input into the linear quasi-static Equation 5.1.

To obtain the stress estimate, $\hat{\sigma}_j$ and hence force estimate given a known input stress and force, the following steps were completed for each CBSR sample:

1. Determine the most likely sensed region:

   (a) An EIT reconstruction image of a loaded DUT of a particular strain, ε_j, at a steady-state conductance was found. Each element having a change in conductance value, $\Delta\rho$, in units mS.

(b) Complete threshold percentage mask on image to localise the sensed region blob(s).

(c) The centre of mass error, E_{CoM}, between each blob and the actual force application CoM was calculated. The blob domain with the smallest E_{CoM}, Ω_s, is chosen for the following steps.

2. Equation 5.1 was rearranged to get an original conductance estimate for each element:

$$\rho_0 = \frac{\Delta\rho}{\alpha\sigma + \beta} \tag{5.7}$$

3. The mean ρ_0 and standard deviation of all of the elements in Ω_s were found.

4. Steps 1 to 3 were repeated for each strain, ε_j, applied and the mean, $\rho_0(\varepsilon_j)$ and standard deviation were calculated.

5. The mean of all $\rho_0(\varepsilon_j)$ across all strain values (i.e. $\varepsilon_j = 5, 10, 15, 20, 35, 30\%$) was calculated as $\bar{\rho}_0$.

6. $\bar{\rho}_0$ was then substituted into Equation 5.1 as ρ_0, which was rearranged for stress to obtain the stress estimate as a function of mean change in conductance, $\hat{\sigma}_j(\Delta\bar{\rho}_j)$, of the sensed blob domain, Ω_s.

5.3 Results

In the following Sections 5.3.1.1-5.3.3.2 the EIT image pre-processing, spatial, temporal, and localised force sensing performance metrics are displayed and quantified. First the steady state electrical noise, σ_n, and noise figure, NF, were determined, as given in Table 5.2. The NF value being a common metric showing noise amplification as a consequence of the EIT algorithm as used by Adler et al [166].

TABLE 5.2: DUT noise figure, NF, and noise, σ_n, at steady-state

CB wt%	**NF**	σ_n [mS]
8	1.20 ± 0.17	0.69
9	1.15 ± 0.11	0.48

5.3.1 1D Material Characterisation

To generate a 2D pressure map from the EIT reconstructions a 1D material electromechanical characterisation was required. The 1D electromechanical relationship can then be extended to form an electromechanical relationship in 2D.

5.3.1.1 Quasi-static Piezoresistivity

Given known strain input data, and measured stress and conductance change output data, a fit is shown in Figure 5.5. The gradient of the linear fit, i.e. the gauge factor, for the CBSR 8 wt% and 9 wt% was calculated as 0.6 and 0.2 respectively. Note that in the 9 wt% relative conductance data the standard deviation of the 5% strain data was similar to that of the mean, hence the linear range of 10 - 30% strain was considered when fitting the curve.

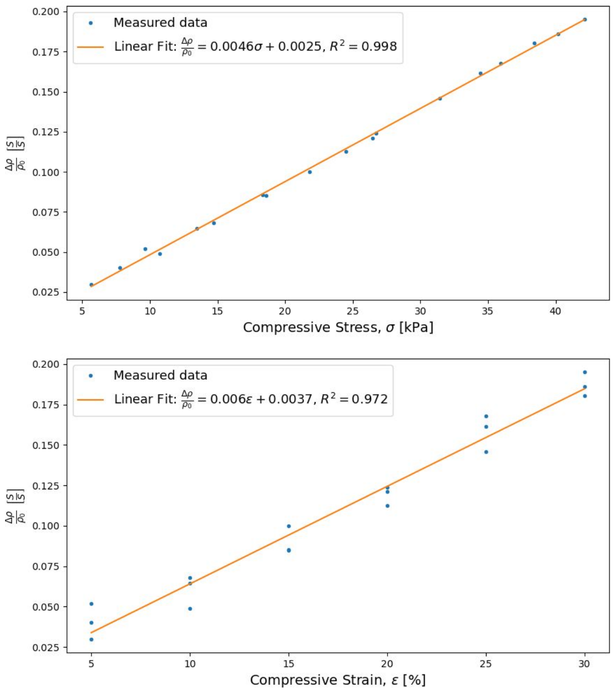


FIGURE 5.5: Conductance change vs. stress (top) and strain (bottom) data and fitted curves for 8 wt% CBSR.

5.3.1.2 Transient Piezoresistivity

The transient piezoresistive effects observed within a PNEC limit the frequency response of the sensor. An example of the transient response of the material to a repeated compressive strain pulse input is displayed in Figure 5.6, clearly showing the stress relaxation of the material due to its viscoelasticity. In Figure 5.7 a loading event is shown with the related stress and resistance relaxation curves. The unloading event similarly has a relaxation period for both stress and resistance in the loading case. Unlike the loading stress transient, the resistance transient has a spike during the unloading relaxation event seen in Figure 5.7. This rising edge and peak of this spike are ignored and the resistance relaxation edge is characterised. For each stress relaxation Equation 5.3 was fitted to the data. Analogously the same was done for the resistive relaxation events observed.

The fitted parameters were found for both 8 and 9 wt% CBSR samples, giving an indication of t

The settling times of the resistance relaxations give an indication of the frequency response of material. Thus, parameters were fitted to a series of relaxations using Equation 5.3, then a series of fit parameters could be used to determine a mean fit. The mean fit was then used to determine the mean settling time over each ten loading events and

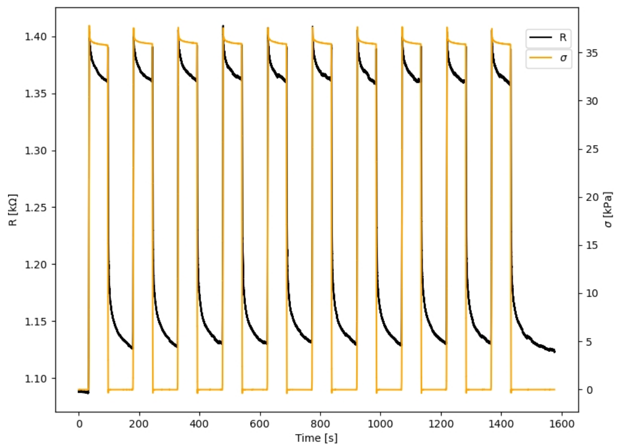


FIGURE 5.6: Compressive loading applied to the CBSR 8 wt% DUT for 10 loading events of 25% strain.

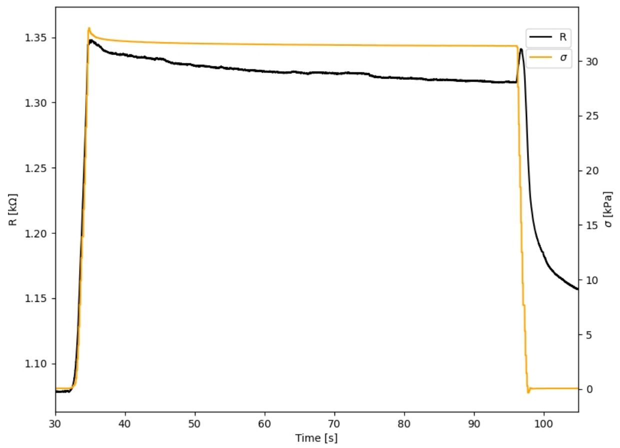


FIGURE 5.7: Compressive loading and unloading transients for CBSR 8 wt% material undergoing a 20% strain pulse from the first pulse given in Figure 5.6.

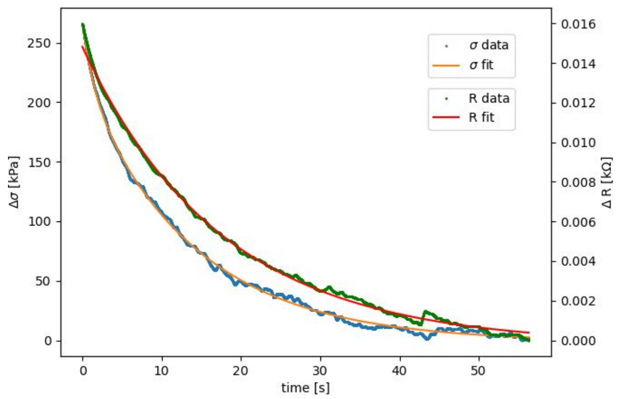


FIGURE 5.8: The fourth 1D load event, L_3 on the CBSR 8 wt% sample using 5% strain showing a stress, σ and resistance, R, relaxation event and their corresponding fitted curves.

each of the six strain values. The mean relaxation settling times were compared for each CB weight percentage as shown in Appendices **??** and **??**.

5.3.2 Sensor Performance Metrics

To validate this 2D pressure sensing platform for specific applications the limits of the sensor must known. Metrics to analyse and quantify the limits, sensor noise and spatial, temporal, and stress performance metrics are given in this section.

5.3.2.1 Pre-processing

The noise floor limits the detection of small forces. First the noise floor was found from the no load steady-state of material. The maximum noise from the first eight frames was found and this maximum was subtracted from all contiguous images in the time series experiment.

After a noise mask has eliminated the steady state noise floor, different percentage thresholds can be used to compensate for different regularisation and different material push area edge softness as shown in Figure 5.9.

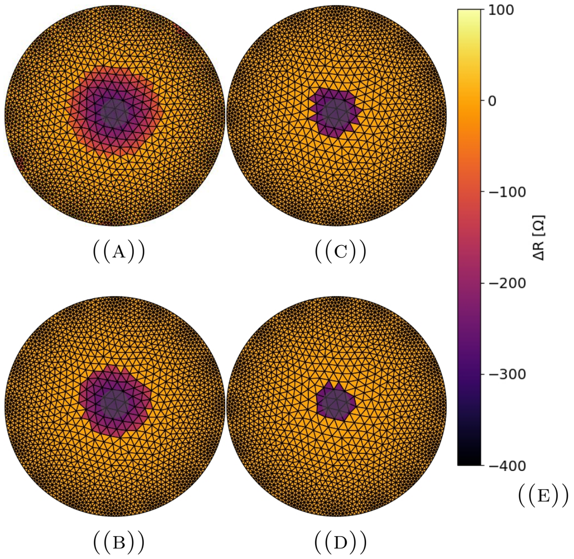


FIGURE 5.9: A series of threshold percentage masks (a) 25%, (b) 50%, (c) 75, and (d) 85% for the same reconstruction given in Figure **??**. (e) is the resistance change scale bar.

The threshold masked image blobs and the force applicator shapes in Figures 5.9(a) - 5.9(d) are compared and quantified in the following section.

5.3.2.2 Spatial Performance

To properly determine where the perimeter of a load is multiple threshold mask filters were applied. To validate the threshold mask percentages the three main performance characteristics were displayed as separate time series for each material, each applied strain, and each threshold percentage mask. These time series show how each metric changed over the course of a loading test sequence and how the metrics vary across the surface of the DUT. An example comparing this time series data for two instances where a 20% strain pulse train was applied to a the nine loading locations with multiple threshold mask percentages is shown in Figures 5.10 and Appendix **??** for for 8 and 9 wt% CBSR samples respectively.

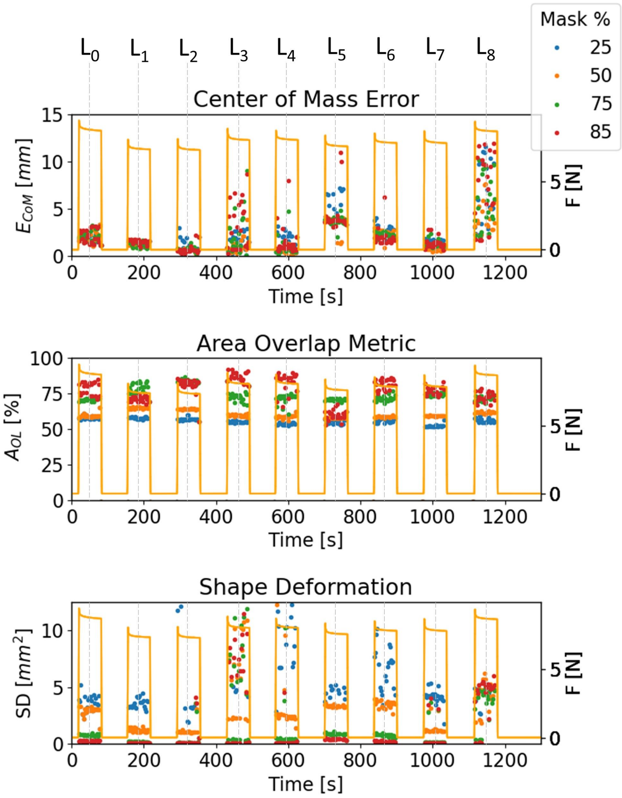


FIGURE 5.10: Spatial performance metrics comparing threshold percentages of 25, 50, 75, and 85% for a 8 wt% CBSR sample being loaded with 20% compressive strain in nine areas, L_{0-8}, shown in Figure 6.7. The force time series plot data is light orange.

The percentage threshold masks from 25 - 85% were compared by finding the mean of all of the spatial performance metrics for each strain from 5 - 30%. The mean and standard deviation for each of these metrics from the data shown in Figure 5.10 is given in Tables 5.3 and 5.4.

TABLE 5.3: CBSR 8 wt% mean and standard deviation for spatial performance metrics across of a nine loads, L_0-L_8 and strain value 20%.

% thresh	E_{CoM} [mm]	A_{OL} [%]	SD [mm^2]
0.25	4.1 ± 6.3	53.3 ± 15.3	10.4 ± 9.3
0.5	3.3 ± 6.3	57.5 ± 16.3	3.6 ± 4.4
0.75	3.8 ± 6.3	69.6 ± 19.8	2.4 ± 5.8
0.85	4.1 ± 6.4	72.5 ± 21.7	2.4 ± 6.5

TABLE 5.4: CBSR 8 wt% mean and standard deviation for spatial performance metrics of nine loads, L_0-L_8, a 85% percentage threshold mask, and strain value 20%.

Load	E_{CoM} [mm]	A_{OL} [%]	SD [mm^2]
L_0	2.05 ± 0.70	80.39 ± 4.19	0.28 ± 0.01
L_1	1.53 ± 0.23	72.32 ± 1.74	0.12 ± 0.01
L_2	0.67 ± 0.41	83.00 ± 2.38	0.48 ± 1.16
L_3	3.29 ± 2.17	87.93 ± 2.68	5.25 ± 3.78
L_4	1.44 ± 1.61	84.81 ± 5.78	2.76 ± 5.65
L_5	4.43 ± 2.09	61.08 ± 3.29	7.39 ± 16.31
L_6	2.03 ± 1.05	82.19 ± 3.55	3.14 ± 8.57
L_7	1.49 ± 0.56	77.68 ± 2.46	0.66 ± 1.25
L_8	6.95 ± 3.67	73.69 ± 3.60	3.78 ± 2.01

5.3.3 Randomised Location and Strain Testing

In a real world application the sensor platform in this work will likely experience a large range of unknown loads in various locations. To ensure that the device operates in a similar fashion to that seen in the structured experimental data, a randomised experiment was completed. The randomised experiment loads were at ten randomised radii, r_{rand}, angles, θ_{rand}, and strain values, within the ranges, 0 - 40% of the domain radius, 0 - 360°, and 5 - 30% respectively.

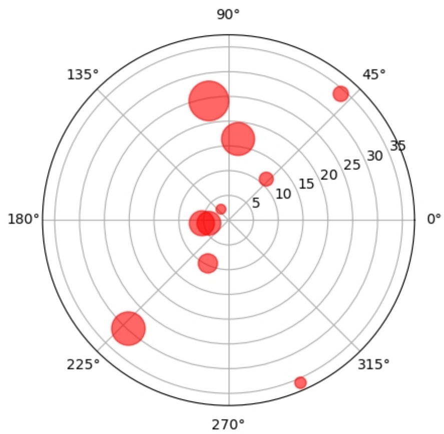


FIGURE 5.11: The ten random load point locations, L_{rand} and random strain values proportional to rec circle size as shown on a polar plot.

Spatial performance metrics for these tests are given in Tables 5.5 and **??** with the load points and their magnitudes shown diagrammatically in Figure 5.11. A pseudo-random number generator with a uniform distribution was used for all randomly generated data.

TABLE 5.5: CBSR 8 and 9 wt% mean and standard deviation for E_{CoM} spatial performance metrics of ten random locations, L_{rand} and random strains ε

L_{rand} (r, θ) [mm, °]	ε [%]	8wt% E_{CoM} [mm]	9wt% E_{CoM} [mm]
(5.4, 10)	17.9	3.1 ± 0.7	3.1 ± 0.5
(29.8, 337)	24.0	1.9 ± 0.4	5.0 ± 0.6
(4.2, 317)	17.0	7.9 ± 2.2	9.0 ± 1.0
(34.0, 70)	10.9	6.0 ± 3.5	2.9 ± 0.5
(35.8, 137)	8.1	13.9 ± 0.6	7.7 ± 4.6
(9.6, 55)	13.8	6.0 ± 1.0	10.3 ± 4.4
(2.8, 260)	7.0	12.0 ± 8.1	28.7 ± 10.9
(11.2, 114)	10.0	5.1 ± 1.2	12.0 ± 1.9
(24.6, 241)	28.5	2.3 ± 0.2	2.4 ± 0.2
(16.6, 253)	23.6	3.1 ± 0.4	7.1 ± 0.5

5.3.3.1 Temporal Performance

Temporal performance is crucial for time sensitive applications and the settling time of the sensing material domain must be known to apply a quasi-static force model. The fitted stress and resistance relaxation parameters were found for both 8 and 9 wt% CBSR samples, giving an indication of the frequency response of material across all experiments. To ensure a good fit all fits with an R^2 value less than 0.85 were eliminated.

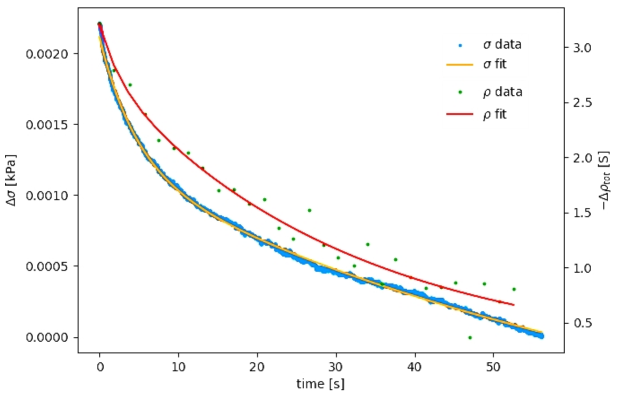


FIGURE 5.12: EIT load event L_0 on the CBSR 9 wt% sample using 30% strain showing a stress, σ and conductance, ρ, relaxation event and their corresponding fitted curves.

The mean settling time for each strain was calculated across relaxations for all strains, all 9 locations, and all 3 trials. The settling times were compared for each CB weight percentage as shown in Appendices **??** and **??**.

5.3.3.2 Localise Force Sensing Performance

To determine the localised force sensing performance the linear quasi-static Equation 5.1 was applied to the percentage threshold masked image blobs developed in section 5.2.3.2.

To determine the force sensing limits of the material, the force estimated erroneously due to the EIT reconstruction noise floor must be determined. The noise floor is the noise observed over a time series of EIT images when the DUT has zero load applied and there are no resistive transient effects present. The noise floor, $\Delta\rho_n$, of unloaded relaxed 8 and 9 wt% CBSR DUT conductance images were calculated as ± 0.33 and ± 0.34 μS respectively. An average DUT inter-electrode conductance, ρ_{int}, of 55.3 and 222.2 μS was derived from Table 5.1 for CBSR 8 and 9 wt% respectively. A relative change of conductance value, $\frac{\Delta\rho_n}{\rho_{int}}$, was then calculated as 5.97×10^{-3} and 1.53×10^{-3} μS for CBSR 8 and 9 wt% respectively. From the quasi-static piezoresistivity Equation 5.1 and the fitted quasi-static piezoresistivity parameters found in Section 5.3.1.1, we calculated the mean force approximation error as 0.17 N for both CBSR 8 and 9 wt%.

The force estimation from the inverse quasi-static Equation 5.1 was compared to the actual force loaded onto the DUT as measured by the force applicator loadcell. Figures 5.13 and 5.14 show data from load applications in the centre (L_0) of the respective 8 and 9wt% CBSR DUTs with a force estimation standard deviation of ± 0.78 and ± 0.81 N respectively.

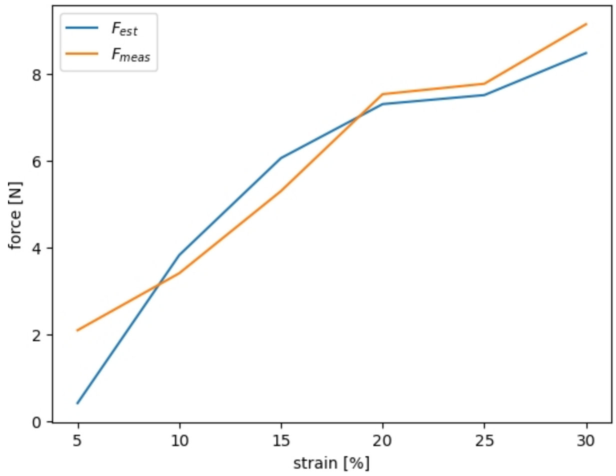


FIGURE 5.13: Comparing force estimates, F_{est}, and actual force measurements, F_{meas}, for 5 - 30% strain centre loading events at L_0 for the EIT sensor system for 8 wt% CBSR

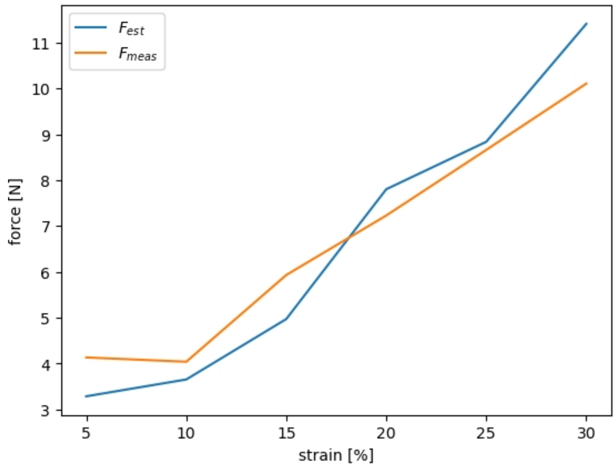


FIGURE 5.14: Comparing force estimates, F_{est}, and actual force measurements, F_{meas}, for 5 - 30% strain centre loading events at L_0 for the EIT sensor system for 9 wt% CBSR

5.4 Discussion

Potential applications that emulate human-like skin pressure sensing characteristics require a forms of quantification to compare the technology to the specific requirements. This work quantitatively characterises performance metrics to help facilitate that comparison and optimisation process. The sensor developed could be likened to slow acting mechanoreceptors within human skin, such as Meissner's corpuscles and Merkel's discs, which combined can detect static pressure, and high resolution touch. For other EIT-based pressure mapping applications to be realised, the metrics developed in this work are some of the core metrics required to determine which soft sensing domains are suitable and are their limits.

5.4.1 Quasi-static Piezoresistivity

To make a low-frequency response load sensor, a quasi-static piezoresistive linear model was created as shown in Section 5.3.1. However, this model is only valid for sufficiently slow pressure applications or after a sufficiently long time period. This time period is determined by the largest expected steady-state relaxation time for the material shown in 5.3.1.1

5.4.2 Pre-processing

The two steps of a noise threshold mask and a percentage threshold mask successfully filtered noise and EIT reconstruction related noise artefacts. The favoured percentage threshold mask chosen for further metrics testing was 85% as this gave the lowest average E_{CoM} and *SD* values from the across all strains applied across all nine loading points.

In the experiments often a blob detection from a previous load will be present in a subsequent load, as expected due to the resistive relaxation. Feature detection could be added in future to ensure that only transients similar to those seen in the initial formation of a blob would signify that the blob is to be analysed. Concurrently, each blob could be tracked individually to determine whether it is a noise artefact or an actual sensed region depending on it's behaviour.

5.4.3 Performance Metrics

To develop sensing domains for future applications, the sensing domains may need take into account certain prior information about the limits of the system.

For example, human hands and feet have some of the highest density of mechanoreceptors in the body. Lower density regions of mechanoreceptors in humans include the back and chest [153]. Higher spatial resolution is require for emulating the pressure mapping of a human hands and feet, compared to the human back and chest. However, the pressure sensing range required by the human hands may be lower than that required by the human feet.

Using this prior information, we can validate the appropriate sensing domain characteristics that give a suitable performance for each different application.

Depending on the application of the sensor the importance of each temporal, spatial, and force sensing performance metrics could all vary.

5.4.3.1 Spatial Performance

All spatial performance metrics, E_{CoM}, A_{OL}, and *SD* are key indicators of whether a loading event has been localised correctly.

The A_{OL} gives a value out of 100 for a certain detected blob. This value is penalised for false positive and true negative elements that overlap (or not) with the force applicator area.

It is important to note, when a force is applied in an small area of a domain, however a blob has been detected over the majority area of the domain, a A_{OL} value of $\leq 50\%$ will be given although the blob detection could be completely false. Although the detected blob and force applicator are 100% overlapping the amount of false positive (i.e. blob elements not overlapping with force applicator area) could cover the rest of the DUT, potentially giving a value nearer to 50% than 0%. From this it must be recognised that this metric does not represent a linear relationship between A_{OL} and the quality of the reconstruction. So the scale of the A_{OL} value to quality relationship was determined empirically as:

$$0 \leq A_{OL} \leq 50\% = \text{Likely Poor}$$

$$50 \leq A_{OL} \leq 70\% = \text{Ok}$$

$$70 \leq A_{OL} \leq 100\% = \text{Good}$$

The *SD* is the mean square error between the force applicator perimeter and sensed region perimeter taken radially from the force applicator centroid, so will likely be lower with a low E_{CoM} and a higher A_{OL}. The closer the *SD* value is to zero the more accurately the shape of the load area applied has been sensed. The *SD* metric is also affected significantly by the quantisation error depending on the mesh coarseness.

Comparing the different percentage threshold masks for the experiment shown in Figure 5.10, it was determined that each percentage mask of 50%, 75%, and 85% gave showed the spatial performance for the E_{CoM}, *SD*, and A_{OL}. However, the standard deviation of these values is comparable to the mean itself therefore looking at the mean performance metric value in each location was shown in Table 5.4. The lowest E_{CoM} was found to be 0.67 ± 0.41 mm, at L_2. The highest A_{OL} value was found to be $87.93 \pm 2.68\%$. The lowest *SD* value was found to be 0.12 ± 0.01 mm^2 at L_1.

The CBSR 8 wt% samples gave better performance metric results than the 9 wt% samples due to the residual transient effects of previous load events as exemplified in Figure **??**. This will be mitigated in future by using a blob separation algorithm whereby each sensed-region/blob is given a weighting based on its appearance time, size, decay characteristic, and performance metric values.

The spatial performance metrics are useful for quantifying future testing with irregular load application area shapes and multiple loading events in future testing to validate a variety of irregular and multi-load test cases. Performance metric inconsistencies in the different load locations show that the electro-mechanical characteristics of the material is varies throughout the material. These metrics would all contribute toward a calibration step to compensate for material inhomogeneity, allowing for a range of materials to be used for the sensing domain.

5.4.3.2 Temporal Performance

Many applications require a minimum frequency response hence a temporal study was completed to characterise the transient effects limiting the speed of the sensor. The study focused on the settling time of transient piezoresistive events in the material for varying strain step inputs. With known PNEC material settling times, a filter could be applied to the output of this sensor to get an estimate of the load applied to the material.

To aid future inverse modelling and use of PNECs as pressure sensor it is important to understand each transient states of a load, including the loading phase, steady state, or unloading phase. It was found that on average that unloading events had a higher settling time than loading transients for both CBSR 8 and 9 wt% composites across all strains tested from 5 to 30%. No clear correlation was found between the settling time of the transient strain events and the strain percentage applied to the material. Mean settling times ranging 29 - 36 s and 29 - 41 s have been observed for the CBSR 8 and 9 wt% composites respectively.

A different sensing region material could provide a higher frequency response, such as a carbon nanotube silicone composite which has shown a lower settling time in previous works [190, 191]. Due to the viscoelasticity and elastic rebound in the material the resistance relaxation from predecessing load applications was often be present in subsequent load events, altering the observed resistance relaxation response. Future algorithms developed would aim to eliminate these predecessing residual relaxations.

Often soft materials are inherently viscoelastic like much soft tissue within the human body [41], so if soft sensor domains are required with a high frequency response this viscoelasticity will need to be compensated for using this work's performance metrics as a foundation.

It is important to note that if the homogeneity in the material is highly irregular, regions of the material will have different degrees of piezoresistivity the frequency response of the material is likely to vary considerably. Further research is required into how the different CB wt % values effect the temporal response of the material.

5.4.3.3 Localised Force Sensing Performance

The sensor platform gave stress estimates that correlated well with the real stress applied to the material, as seen in Figures 5.13 and 5.14. These stress estimates were gathered from the steady-state data gathered from the EIT measurements at approximately 1.5x the settling times found in Section 5.3.3.1 using the algorithm given in Section 5.3.3.2 to ensure the data was at steady state.

Stress relaxation of the composite CBSR material as a whole gives a good indication of macro-mechanical behaviour of the CBSR. It was postulated that the resistance relaxation gives an enhanced insight into the micro(and nano)-structural behaviour of the CBSR composite, because of the different observed behaviours of the CBSR stress and resistive relaxation and also how these relate to different CB weight percentages and their dispersion.

5.4.4 Real World Applications, Manufacturability, and Scalability

Using EIT-based pressure mapping on a larger scale is feasibly as shown by the use of ERT in geophysics [192]. Potential larger-scale applications include adding a pressure mapping layer under a tennis court to map force exerted by athletes onto a court or a method of measuring foot traffic in buildings and urban areas. The use of the performance metrics discussed in this work would be applicable for both scenarios.

For the tennis court application, the importance of player location and speed may be more important than detecting the footprint shape and exact force applied to the court surface. This means that the E_{CoM} and decay time values would be more heavily weighted than the *SD* and force values, and hence could be tuned for these characteristics. For the urban floor mat application, the importance of footprint shape and force estimation may give useful insight into the physical demographic of people or animals walking across the mat [193]. This may mean that the *SD* and force resolution values are more highly weighted in the design and process.

Larger-scale applications of an EIT-based sensor come with challenges such as, scaling the electronics driving the EIT measurement, fabricating such a large homogeneously piezoresistive domain, and ensuring the reliability in a range of outdoor environmental conditions. Smaller-scale applications are limited by the conductive particle size in the

PNEC. A sensing domain thickness sufficiently larger than the average agglomerate size would be required for reliable EIT mapping and force estimates.

Various forms of tribological wear on the device sensing region would alter the piezoresisitive characteristic of the device. Encapsulation of the device could be implemented to minimise wear and increase hermeticity.

The most obvious limitation of this sensor is the frequency response of the material as shown in Figures **??** - **??**, which could be algorithmically filtered or inverse-modeled to be corrected. Else, other more responsive, less viscoelastic materials could be used, and/or a capacitive EIT-based pressure mapping device used to improve the frequency response of this device. Otherwise the use of a time series dependent neural network, such as an LTSM, RNN, could be used to inversely model such events.

Mass production of an EIT-based sensor would use the performance metrics given in this work to calibrate and quality-check the sensing domain and boundary electrode connections. This work also found that using pin boundary electrodes adds to the durability and stability of electrical connection in this device.

5.5 Summary and Conclusions

An EIT-based piezoresistive sensor using a custom made carbon black silicone rubber composite material has been developed for sensing compressive pressure events and applying performance metrics to obtain the validity of the output EIT images. To be able to apply this EIT-based PNEC pressure sensor to a variety of scenarios, replacing human-like touch, performance metrics has been formed to quantify the sensor's suitability for each application. Sensing domains of 8 and 9 wt% carbon black silicone rubber have been tested using: 6 strain values, 9 load locations, and 3 trials. From this raw data we have calculated data for: spatial resolution, transient settling time, and force sensor resolution.

It was shown that the CBSR 8 wt% sample out performed the CBSR 8wt% sample in terms of spatial and temporal metrics across a range of experiments. The best performance metrics observed in the CBSR 8 wt% sample for E_{CoM}, A_{OL}, and SD, were 0.67 $\pm$ 0.41 mm, 87.93 $\pm$ 2.68%, and 0.12 $\pm$ 0.01 mm^2 respectively for three different load locations. For the sensor domains tested, average settling times of between 19.0 - 44.5 s and 22.5 - 36.0 s were determined for 8 and 9 wt% CBSR samples. A quasi-static conductance-force model of the material was developed with an accuracy of $\pm$ 0.78 and $\pm$ 0.81 N for a range of strains from a centre load test for 8 and 9 wt% CBSR respectively.

Using these performance metric data in future work a piezoresistively inhomogeneous sensor domain could be, calibrated to homogenise the apparent domain piezoresistivity, compensated for transient phenomena, and sense loads with a known degree of accuracy. All of these factors contribute to optimising the EIT-based 2D pressure mapping sensor for different applications. Future work also includes the development of a low-cost, small circuit to capture the the data discretely to open up a larger range of applications. The work shows promise for future use of an EIT-based sensor in a variety of applications requiring a soft sensing domain and non-invasive rigid electrodes.

Chapter 6

Giving Artificial Muscles the Sense of Touch

The content from this chapter is predominantly from the manuscript published in the proceedings of Electroactive Polymers Actuators and Devices XXVI.

ABSTRACT

Dielectric elastomer actuators commonly use flexible conductive electrodes to apply an electric potential to the system. Depending on the material used, these electrodes often possess predictable piezo-resistive properties. Combining electrical impedance tomography (EIT) with a dielectric elastomer actuator (DEA) is investigated in this work to map compressive forces occurring throughout the electrode surfaces. This technology could allow for enhanced closed-loop control of electroactive actuators, extending their already extensive set of applications. This deformation mapping system also has the potential to be used with other piezoresistive materials, opening up more applications requiring a large hardness range and pressure sensitivity. With the material used in this work, the DEA-EIT device has an inherent trade-off between actuation and pressure mapping accuracy driven by the compliant electrode thickness of the DEA. The DEA-EIT device exhibited actuation strains of 2.5 % with a mean centre-of-mass error from a range of loads applied were 7.9 $\pm$ 0.7 mm for 2 mm thick DEA electrodes. It is proposed that future work on custom hardware could be devised for the DEA-EIT system so the sensing and actuation can occur concurrently in real-time. Real-time control mean that applications requiring human-like manipulation can be designed, ranging from biomedical implant devices to agricultural processing equipment.

6.1 INTRODUCTION

Fine motor manipulation, pressure sensitivity, and pressure mapping are some core attributes of skin and muscle tissues when innervated to the brain. These functions can be emulated and combined with two core technologies, Dielectric Elastomer Actuators (DEAs), and Electrical Impedance Tomography (EIT) based pressure mapping.

DEAs have been used to mimic biological muscles in many applications, because of the technology's likeness to biological muscle in terms of elasticity, energy density, and various potential shapes/topologies [194–196] . In previous research, it was determined that pressure mapping similar to that of human mechanoreceptors could be emulated using EIT with a piezoresistive nanoparticle elastomer composite (PNEC) in a planar sheet format[20] . The key qualities of the EIT-based sensing platform were that pressure estimates could be obtained, the pressure could be mapped, and the spatial performance could be quantified[21] . Like DEAs, this sensing technology has a likeness to human tissue in terms of mechanical characteristics such as elasticity, and the potential of various topologies.

Alongside the visual and other sensory feedback, animals receive when actuating muscle tissue, pressure-sensitive mechanoreceptors are present within the muscle tissue and soft skin tissue to aid control the extent of a muscle contraction. This forms a multi-sensor closed-loop control system with a complex biological control regime. This work is looking towards creating a closed-loop control system which utilises a DEA-based artificial muscle and an EIT-based artificial skin.

6.1.1 Background

The fundamental principles and a brief explanation of the state-of-the-art of each DEA and EIT-based sensor technologies are given in this section. A review of pressure mapping devices with actuation capabilities was then completed. At the time of completing this work, no literature had been found regarding the combination of these two technologies using PNEC electrodes on a DEA for successive execution of sensing and actuation events.

6.1.1.1 Dielectric Elastomer Actuators

DEAs are often referred to as artificial muscles because they share similar characteristics to biological muscle. DEAs have been proven to produce strains larger than 1600 %[83] which is significantly larger than that of regular biological muscle. However, large DEA strains can often be at the cost of actuator instability and a low effective force. DEAs have a high work and power density comparable to that of biological muscle and have been found experimentally to have energy densities of around 3.4 $J.g^{-1}$ and theoretically an order of magnitude more[134, 197] . A dielectric elastomer actuator (DEA) is a form of soft robotic actuator that induces deformation with an applied electric field. Although commonly used as an actuator, this technology offers versatile applications as an energy generator[197–199] or sensor and provides attractive features such as high energy density, large displacements, and fast response times. A simple configuration of a DEA is a circular parallel plate capacitor, which consists of a thin elastomer sheet between two compliant conductive electrodes, as shown in Figure 6.1. When a voltage is applied to the compliant electrodes, the electrostatic force arises between the electrodes causing the dielectric elastomer (DE) membrane to contract by a decrease in thickness and an increase in area. The resulting actuation is controlled by changing the applied voltage. The region encompassing the two compliant electrodes and the DE portion sandwiched between them is called the 'active region'. For a simple DEA such as the one shown in Figure 6.1, a simplified formula for the electrode static force on the parallel plates is given in Equation 6.1.

$$\sigma_{es} = \varepsilon_0 \varepsilon_r \frac{V^2}{z_{de}^2} \tag{6.1}$$

Where σ_{es} is the electrostatic stress, V is the applied voltage, z_{de} is the DE thickness, ε_0 is the permittivity of free space, and ε_r is the relative permittivity constant of the DE, which is a function of strain[200–202] and applied voltage[36] . This can be expanded to include the bulk modulus, K, of the DE as given in Equation 6.2.

$$S_{z_{de}} = \frac{\sigma_{es}}{K} \tag{6.2}$$

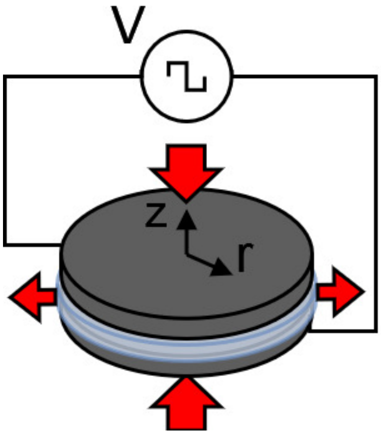


FIGURE 6.1: A circular DEA exemplifying its actuation principle.

6.1.1.2 Soft EIT-based Pressure Mapping

A soft EIT-based pressure mapping sensor has the ability to estimate the magnitude and location of force loads in a planar PNEC material. The hardware required usually consists of a piezoresistive sensor domain with attached boundary electrodes, EIT driver electronics, and a reconstruction processor. Boundary electrodes allow a non-invasive method of pressure mapping without compromising a monolithic piezoresistive material. Several researchers have created an EIT-based pressure mapping sensor using a range of piezoresistive domains and custom or lab-based hardware [21, 106, 158, 179–181].

EIT devices can perform at image frame rates higher than 50 Hz, however using EIT to map and quantify piezoresistive pressure events has the potential to give a faster sample rate due to the use of DC instead of AC. Due to the viscoelastic and resistive nature of the sensor, the frequency response can be lower than that of the frame rate of the sensor depending on the piezoresistive sensing domain used.

The stages required to generate a pressure image using EIT can be simplified into three core stages,

1. Data acquisition
2. Image reconstruction

3. Inverse pressure model

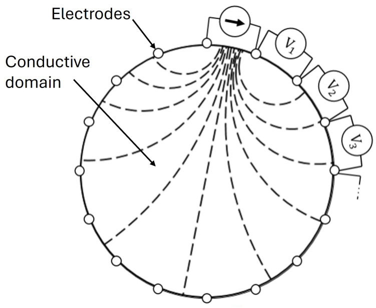


FIGURE 6.2: A 16 electrode circular EIT domain setup exemplifying its electrical function. Where the dashed lines are representative of an applied electric field[20] .

Data acquisition involves an excitation drive pattern to be applied to the piezoresistive sensing domain, which consists of the injection of a known current or voltage through two boundary electrodes connected to the material domain as shown in Figure 6.2. Typically an adjacent electrode drive pattern is used in literature as it is used in this work[180] . Concurrently all voltages at the other boundary electrodes of the material domain are read. Then the known current (or voltage) source is applied to the next set of adjacent electrodes, and all of the other adjacent electrode voltages are read once more. This process is repeated until it has been deemed there have been sufficient readings to solve the inverse problem and generate a conductance, ρ, distribution estimate of the material domain. Finally, an inverse model converting the conductance estimate of the material domain can be converted into a pressure map using an inverse model.

6.1.2 Simultaneous Pressure Mapping and Actuation

Various researchers have demonstrated and proposed the use of self-sensing DEAs for closed-loop control looking at the one-dimensional deformation of a DEA using capacitance[35, 37, 38, 203] . Multi-degree-of-freedom (multi-DOF) DEA topologies have been created by several researchers [204–207], allowing for a broader range of applications. The complex actuation mechanisms discussed in these papers give rise to the question of having more resolute sensor data for such topologies to aid with the control of such multi-DOF devices.

To ensure the DEA maintains minimal change to the parallel plate topology, the compliant piezoresistive electrodes can be used and/or altered to be able to determine the deformation of the DEA in more than one dimension using EIT. EIT-based DEA sensing offers several potential advantages over other methods of pressure sensing, such as the use of boundary electrodes, using existing compliant electrode material as sensing domains, simple drive circuitry, and various sensing domain shapes.

6.2 METHODOLOGY

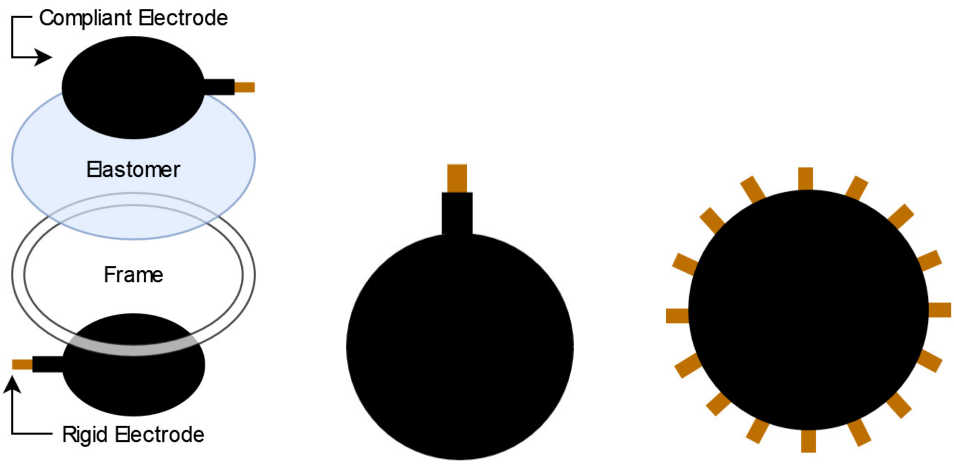
The DEA-EIT actuator-sensor-hybrid system required the two technologies to be verified and fabricated individually before being integrated to observe the effects of combining the two technologies relative to their independent forms. The following sections discuss the fabrication process for the DEA and EIT systems and the integration of them both into a DEA-EIT system.

To optimise the actuation and sensing capabilities of the DEA-EIT system different parameters of design were altered, such as the compliant electrode composite, DE, circumferential electrodes, and magnitude of DE pre-stretch and sizing. The methodology explores the reasoning behind certain design choices for the fabrication of the DEA design seen in Figure 6.3.

6.2.1 Dielectric Elastomer Preparation

The fabrication of the DEA used a rigid acrylic frame to attach the pre-stretched elastomer. For simplicity, a circular frame was chosen with the DE at a radial pre-stretch of +10%, i.e. $\lambda_r = 1.1$, as this is well within the DE's more predictable linear elastic region. The circular acrylic frame of 178 mm inner diameter was fabricated from laser cut acrylic of 4 mm thickness to ensure rigidity.

To achieve uniform stretch of the elastomer sheet, a toroidal shower hose mechanism was placed on the relaxed sheet of 4910 VHB tape (3M, Saint Paul, USA), which would act as a pre-stretcher annulus. The toroidal mechanism has an axis of rotation along its circumference as shown in, giving the ability to roll and stretch the elastomer equiaxially to the desired pre-stretch.



((A)) Circular DEA construction

((B)) Single circumferential electrode

((C)) Multiple circumferential electrodes

FIGURE 6.3: Mechanical fabrication of the circular DEA-EIT platforms

6.2.2 Compliant Electrode Fabrication

Compliant electrodes (or active area) were fabricated using acrylic moulds of varying dimensions. Three thicknesses, z_{ce}, of the compliant electrode were fabricated, 0.5 mm, 1 mm and 2 mm, with two circular compliant electrodes of 100 mm diameter. Different thicknesses were explored as this would vary the actuation and sensing performance of the electrodes.

Two compliant electrode mediums were used in this work, carbon black (CB) powder and a carbon black silicone rubber (CBSR) composite. Compliant electrodes solely made of CB powder have been used in DEAs in previous literature[208–210] , hence this work uses the same compliant electrode type to generate reference data. The CB powder was used to make a single circumferential electrode configuration DEA as a reference to compare to the following DEA-EIT experiments. The CBSR composite was used to make both single (Figure 6.3(a)) and multiple (Figure 6.3(c)) circumferential electrode configurations of DEAs. The CB powder used in all of the compliant electrode samples was Vulcan XC-72 powder (Fuel Cell Store, Bryan, USA). The CBSR composite had 8% CB by weight mixed with DragonSkin 10NV silicone rubber (Smooth-On, Macungie, USA). This composite is a piezoresistive medium that has proven useful for EIT pressure mapping and sensing in previous work[21] , and DEA actuation[208, 209] .

Using the liquid silicone rubber, the CBSR composite mixture was formed by combining part A of the silicone solution and 8 wt% CB and mixing by hand for 10 s. The mixture was then placed in the ARV-310PCE planetary vacuum mixer (Thinky Inc., Tokyo, Japan) to complete a mixing cycle with 500 RPM for 45 s followed by a cycle with 800 RPM for 45 s. In the same mixing container, part B of the silicone solution was added to the mixture and stirred by hand for 10 s and immediately the same mixing cycle in the planetary mixer was completed again. After the cycle was completed, the composite was poured into the mould with attached circumferential copper tape electrodes. The CBSR mixture was then placed in an oven at 80 °C for 4.5 h to ensure the composite was sufficiently cross-linked.

Two types of compliant electrode configuration have been fabricated in this work, single circumferential electrode and multiple circumferential electrodes. The single circumferential electrode configuration was purely for testing the DEA actuation. The multiple circumferential electrode configuration consisted of 16 evenly spaced circumferential electrodes. The multiple circumferential electrode configuration was for testing both pressure mapping and actuation functionality of the DEA. Prior to curing the compliant electrode in a circular mold, the conductive copper tape circumferential electrodes were placed into the mold. The width of the circumferential electrodes was 8 mm. The circumferential electrodes were placed with a 3 x 8 mm area embedded in the compliant electrode circumference edge with the rest of the circumferential electrode protruding for easy access to external electrical connections.

6.2.3 DEA Hardware

The excitation voltage for the DEA was provided by a Trek 610E high voltage supply (Advanced Energy Industries, Fort Collins, USA) providing a maximum voltage output of 10 kV DC. The DEA was placed in a clear insulated box with the high voltage supply leads attached to the rigid copper electrodes of the DEA. An iPhone 11 camera (Apple

Inc., Cupertino, USA) was used to obtain images of the radial compliant electrode strain as shown in Figure 6.4. The current limit of the high voltage supply was set to its maximum of 2 mA to ensure the DEA actuation was not limited by the charging of the compliant electrodes.

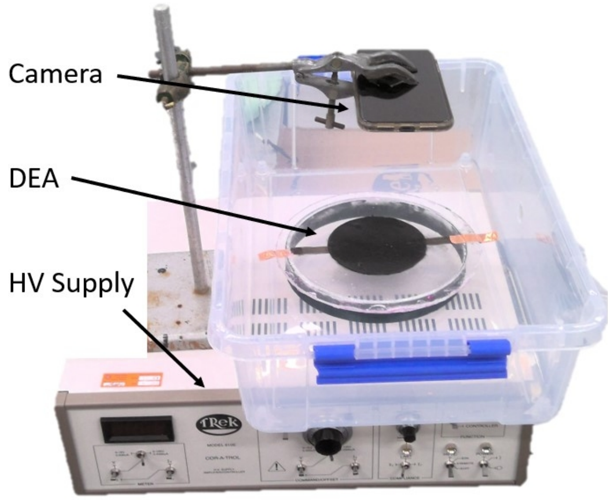


FIGURE 6.4: DEA excitation and measurement setup

6.2.4 EIT Hardware

The EIT hardware allows for data collection to reconstruct a conductance map of the of piezoresistive composite used as compliant electrodes in the DEAs. The hardware required for this function, shown diagrammatically in Figure 6.5, includes a 2634b source measure unit (SMU) (Keithley, Solon, USA), a custom 4:16 multiplexer (MUX) PCB, an ESP32 development board, a Cartesian force applicator, and a reconstruction and control computer.

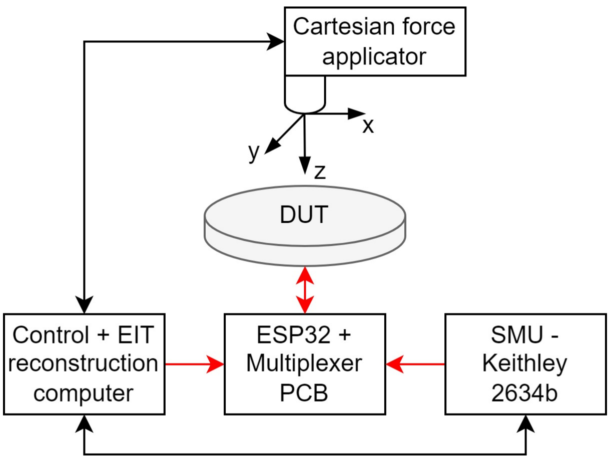


FIGURE 6.5: Architecture of the force applicator and EIT measurement system [21]

The SMU provides a constant current value of 1 mA and reads a series of voltages through the MUX PCB required for the EIT drive pattern. An adjacent electrode EIT drive pattern was used for EIT through the circumferential electrodes of the compliant electrode. The MUX PCB and SMU and controlled by the control computer. Once the data for each image reconstruction has been gathered the control computer was also used for the reconstruction of the conductivity maps of the compliant electrodes. The Cartesian force applicator is made up of a MK3s 3D printer (Prusa, Prague, Czechia) integrated with a loadcell and fabricated applicator tip to apply loads and hence generate data for pressure magnitude and localisation quantification.

6.2.5 Experimental Method

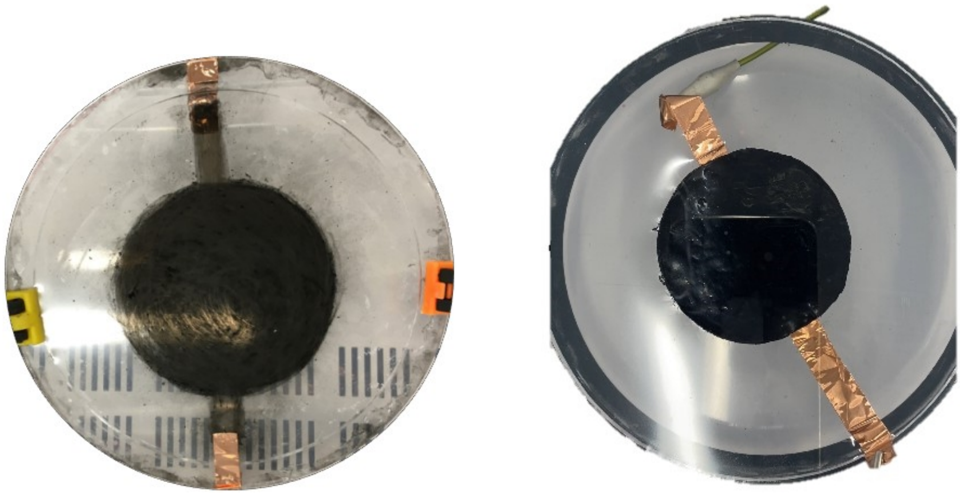
Before attempting to have simultaneous DEA actuation and EIT sensing occur in the same device, each system had to be tested independently. First DEA strain voltage relationships were explored, followed by EIT-based pressure mapping of the DEA electrodes. Finally, the same samples used for EIT testing were subsequently integrated into a DEA device for actuation testing.

6.2.5.1 DEA Validation

DEA actuation strain measurements were taken from voltages 5 kV to 10 kV in 1 kV increments. The SNR of strain measurements of the DEA excited with voltages < 5 kV were deemed to be too low to generate meaningful data. The excitation voltage was toggled between on and off states waiting for the strain deflection to reach a steady state for the strain measurements. Five radial strain images were captured and measured. The measurements were then averaged to minimise error and determine the radial strain uniformity. The radial strain was found by measuring the radial change of the circular compliant electrodes between relaxed and electrically contracted states. From the radial compliant electrode strain the planar and thickness deformation of the DE was estimated. The DE is assumed to be incompressible and have a Poisson's ratio, ν, of 0.5. The adhesion between the compliant electrode and the DE is assumed to be perfect to simplify the model used here. The thickness strain, $S_{z_{de}}$, is calculated using Equations 6.3[208] and 6.1.

$$S_{z_{de}} = \frac{1}{(S_{r_{ce}} + 1)^2} - 1 \tag{6.3}$$

Where S_r is the radial strain measured from the equi-biaxial actuation. The elastic modulus of a hyperelastic material such as VHB is often defined as a non-linear function of strain and strain rate[211] . In this work a bulk modulus value, K, of 142 kPA was used. The bulk modulus was determined by doing a meta-analysis and averaging of the elastic moduli determined at steady state 10% strain for VHB 4905 material in previous literature[38, 211, 212] . The relative permittivity, ε_r, used in Equation 6.1 was approximated to be 4.5 ± 0.2 due to pre-strain effects seen in literature[200–202, 213] .



((A)) The reference DEA made with CB powder compliant electrodes

((B)) A DEA with 0.5 mm thick compliant electrodes at steady state contraction and excited at 10 kV.

FIGURE 6.6: Two main types of DEA compliant electrodes used in this work.

6.2.5.2 EIT-based Pressure Mapping on DEA

A load sequence was devised to ensure that forces in various locations of the compliant DEA electrode could be localised using EIT. A similar test procedure used in previous works[21**?**] was applied to the three individual compliant electrodes thicknesses, z_{ce}, of 0.5, 1, and 2 mm. Nine load application points were determined on the material at three distinct radii as shown in Figure 6.7.

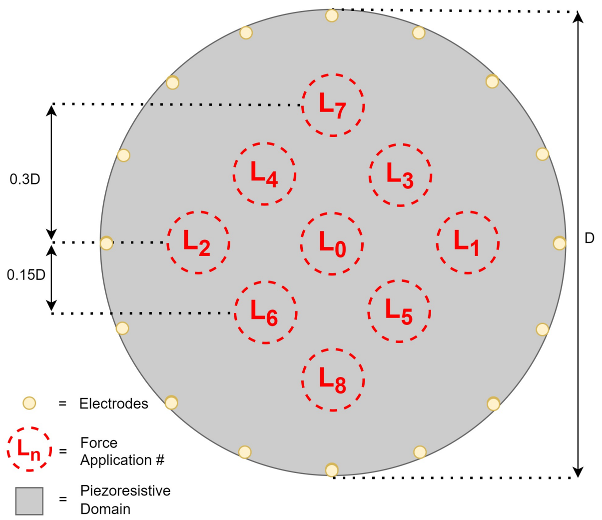
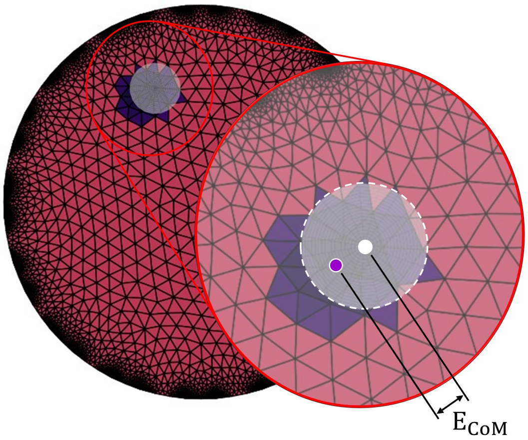


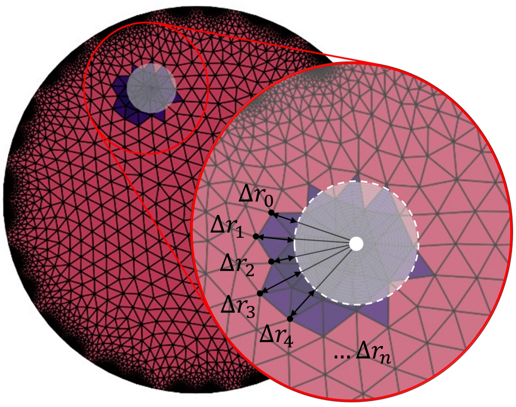
FIGURE 6.7: Load application areas used for compressive stress testing in order of application[21] .

A Cartesian force applicator applied the loads with varying strains in the nine locations. Once the compliant electrodes had undergone the first load application tests, they were applied to each side of a DEA and tested again using the same sequence of nine loads. Compressive strains from 0 to 30% of the electrode thickness in 5% increments were applied to each of the load points using a flat circular 13 mm diameter force applicator. When applying the compressive strain to the compliant electrodes, a strain rate of of 16.67%.s^{-1} was used to minimise piezoresistive transient phenomena. After completing the load tests on individual compliant electrodes the compliant electrodes were attached to the DE surface. Next the load test was completed again with the DEA placed on a flat surface.



Stre

((A)) E_{CoM}



((B)) *SF*

FIGURE 6.8: Spatial performance metrics used for determining the accuracy of a blob as a load area estimate.

$$SF = \left(\sum_{n}^{i} \Delta r_i^2 \right) / n \tag{6.4}$$

After gathering all of the experimental data from applied loads, the data was used to generate EIT conductance images. To form blobs as estimates of the applied loads, post-processing was completed by applying an 85 % threshold mask to the EIT image. These blob images were subsequently analysed using two spatial performance metrics, the centre-of-mass error, E_{CoM}, and the shape fit, *SF*, as exemplified in Figure 6.8. The E_{CoM} values were determined by calculating the difference between the CoM of the actual load and the blob representing the load estimate. The *SF* was determined by calculating the radial mean square error between all of the, n, perimetral nodes of the blob load estimate and the actual load circumference, as taken from the CoM of the actual load area.

To ensure the EIT domain reconstruction was geometrically accurate the circumferential electrodes were modelled in the meshing software to the same width as the real circumferential electrodes. However, the embedded depth of the circumferential electrode was not modelled. Due to the manual nature of the fabrication, significant error of up to 3 mm in the circumferential spacing of the rigid EIT electrodes was present, a factor which would be improved in future iterations.

6.3 RESULTS AND DISCUSSION

A system was created that could both generate strain and map strain events using common DEA componentry with a circular DEA topology. The major limitations of integrating the two technologies were explored and quantified as a starting point for the further optimisation of such a device through the results given in this section.

6.3.1 Fabrication

Fabrication methods were successfully developed to create a range compliant electrode composites for successful DEA actuation and EIT-based pressure mapping. Prior to constructing the DEA, the compliant CBSR inter-electrode resistances, R_{int}, were measured in a similar fashion to voltage shown in Figure 6.2 to ensure the sufficient conductivity and CB dispersion for DEA actuation and EIT-based sensing. The R_{int} values between the adjacent circumferential electrodes for all samples was consistently < 20 kΩ, as shown in Table 6.2. These R_{int} values indicated a sufficiently low resistance for our electrical drive circuitry setup and indicates sufficiently homogeneous CB dispersion. Further validation to check the homogeneity of the whole domain could be done using other invasive or non-invasive methods such as a nail-bed resistance test and/or spectroscopic imaging. Inhomogeneity quantification at a scale smaller than that of the thickness of the material sample is important to obtain a higher SNR and hence *NF* value.

Stress-strain characterisation of DEs in literature clearly shows a hyperelastic softening effect between 50 to 400 % strain for VHB film material, whereby the elastic modulus decreases to 40 - 70 kPa[211] . This is significantly less than the assumed $\sim$ 142 kPa elastic modulus resulting from a 10 % pre-stretch in this work. This hyperelastic region should be determined using both the DE and compliant electrode materials' hyperelastic regions to ensure the K_{eff} is minimised for maximal actuation strain, $S_{z_{de}}$.

6.3.2 DEA Validation

Before testing the piezoresistive compliant electrodes for both actuation and pressure mapping capability, the reference DEA was tested for its voltage actuation strain relationship. The theoretical actuation strain versus voltage was compared to the measured strain for the reference DEA shown in Figure 6.6(a). The actuation strain data gathered is shown in 6.9. The range given was derived from substituting the range of K and ε_r parameters found in previous similar works[38, 211, 212] into Equations 6.1 - 6.3.

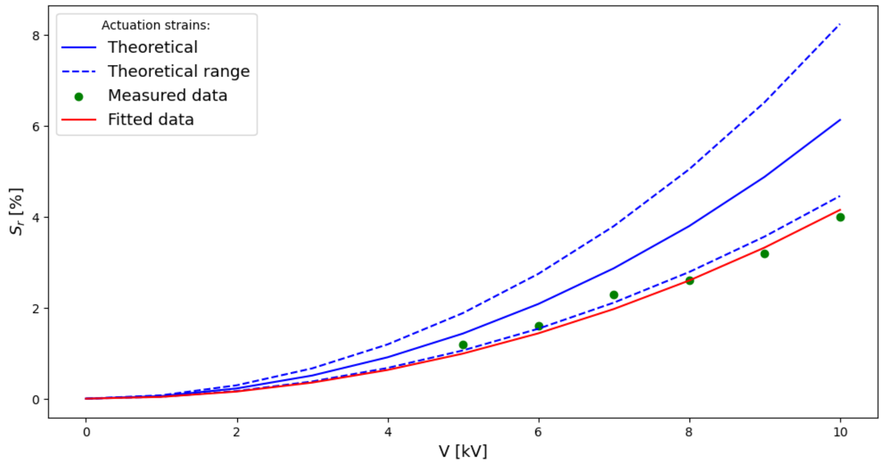


FIGURE 6.9: Comparing the measured CB reference DEA strain, the theoretical strain average and range, and the data fitted to Equation 6.3 by fitting either parameter K or ε_r.

Through actuation testing of different compliant electrodes applied to a DEA, models were fitted to the voltage-strain data gathered. The curve fit shown in Figure 6.9 was found to have a linear set of solutions for K and ε_r with values similar to those limits of the material given characterisation determined in previous literature [214] .

The CBSR compliant electrode experiments showed significantly smaller strains relative to the DEA with the CB powder compliant electrode as displayed in Figure 6.10. The effective mechanical impedance for the DEA with the CBSR compliant electrodes was significantly increased due to the relative thickness of the CBSR electrode and similar bulk modulus relative to the DE VHB material. Hence an effective bulk modulus, K_{eff}, was calculated from fitting to the measured data, as a sum of the existing DE bulk modulus, K, and the effect of the compliant electrode's bulk modulus. K_{eff} is the a key characteristic of using thick compliant electrodes on a DEA that limits the actuation performance. When calculating K_{eff}, ε_r is assumed constant, as the effects of this different compliant electrode thickness on ε_r is assumed relatively negligible.

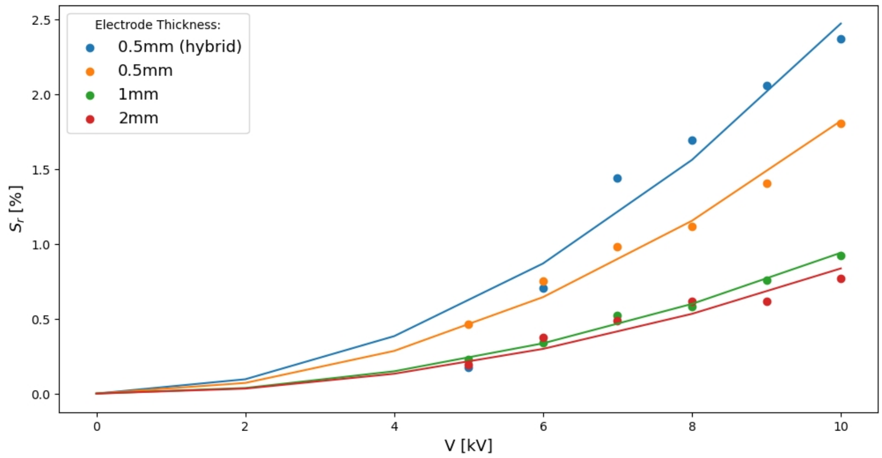


FIGURE 6.10: Comparison of the voltage strain relationships between the 100 mm diameter compliant electrodes of various thicknesses, z_{ce}, used for the DEA.

The effective bulk modulus impeding the actuation of the DE was calculated for each CBSR compliant electrode thickness by fitting to Equation 6.3 with the results displayed in Table 6.1. To further enhance the actuation strain, $S_{z_{de}}$ of the DEA-EIT device, the

TABLE 6.1: Effective bulk modulus, K_{eff}, and coefficient of determination, R^2, for each fit of voltage-strain data for the CBSR compliant electrodes.

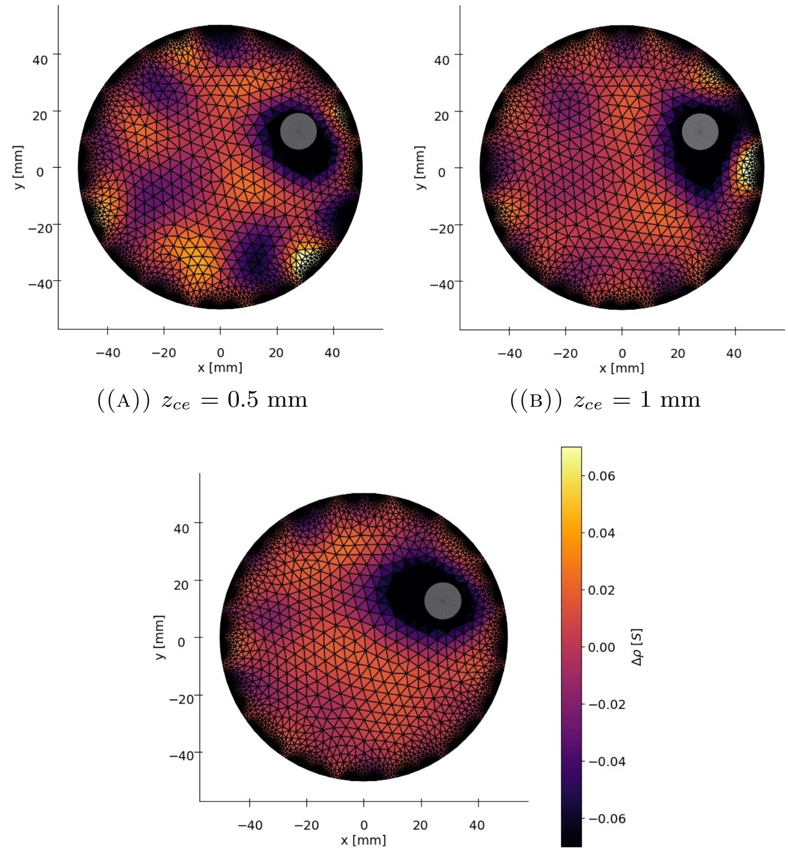
z_{ce} [mm]	K_{eff} [kPa]	R^2
2	966	0.86
1	860	0.99
0.5	450	0.98
0.5 (hybrid)	334	0.91

compliant electrodes were hybridised such that one of the compliant electrodes was made from CB powder and the other from the CBSR composite. The hybridised results for the K_{eff} value are also given in Table 6.1.

It is well known in literature and is intuitive that the mechanical characteristics of a DEA's compliant electrodes have a significant effect on the actuation performance [214, 215] . However, there has been a lack of empirical evidence and subsequent modelling on quantifying how much the thickness of a piezoresistive composite electrode affects actuation performance. This work provides empirical data to begin creating and validating models for thick electrode DEAs, as a step towards creating an objective function to optimise for a DEA for both actuation and sensing performance.

6.3.3 EIT Validation

EIT was used to map nine compressive loads applied throughout the material successfully. Validation of the EIT sensing method on the compliant electrodes was carried out for the three different electrode thickness, z_{ce}, values to see the differences the thickness may have on the pressure mapping characteristics' spatial performance. Figure 6.11 exemplifies the difference in the conductance reconstructions for a load.



((A)) $z_{ce} = 0.5$ mm

((B)) $z_{ce} = 1$ mm

((C)) $z_{ce} = 2$ mm

((D)) Scale

FIGURE 6.11: A 15% strain compression at point L_1 applied to 100mm diameter compliant DEA electrodes of 3 compliant electrode thicknesses.

A significant factor for determining the minimum pressure able to be detected is governed by the noise floor experienced when the domain is in a steady relaxed state. A metric used to quantify the noise floor is the noise figure, *NF*, which is commonly used in other applications[21, 166] of EIT.

A metric used to determine the minimum resistance change measured and hence pressure sensed is the *NF* which can be seen analogously as a SNR. It was found that for increasing values of z_{ce} the *NF* values also increased.

To quantify the domain homogeneity, the inter-electrode resistance, $\bar{R}_{int}$ (of adjacent electrodes) data was gathered alongside the NF, as shown in Table 6.2.

TABLE 6.2: Noise factor, *NF*, domain steady state noise, σ_n and mean inter-electrode resistance, $\bar{R}_{int}$, for each thickness of compliant DEA electrode used for EIT

z_{ce} [mm]	*NF*	σ_n [mS]	$\bar{R}_{int}$ [kΩ]
2	0.99	16.55	4.40 ± 0.69
1	0.98	27.48	7.72 ± 1.14
0.5	0.96	38.86	9.91 ± 2.16

To quantify the localisation performance of the loads applied to the DEA compliant electrode the centre-of-mass error, E_{CoM}, and shape fit, *SF*, of the sensing system were calculated. From three nine load experiments a set of mean E_{CoM} is given in Table 6.3.

Mean values for the spatial performance metrics, E_{CoM} and *SF*, were gathered for each strain, each thickness, and at each load point. Spatial performance metric means from one nine load experiment are given in Table 6.3. The *SF* values are found using Equation 6.4.

TABLE 6.3: Mean E_{CoM} and *SF* values (± std) obtained for each DEA compliant electrode thickness at 20% strains loads

z_{ce} [mm]	$\bar{E}_{CoM}$ [mm]	$\bar{SF}$ [mm^2]
2	4.99 ± 65.2 %	48.86 ± 7.86
1	7.82 ± 90.4 %	49.32 ± 11.79
0.5	21.03 ± 74.2 %	68.74 ± 31.18

6.3.4 Simultaneous Actuation and Mapping

This work has shown that using the same device for both actuation and pressure mapping shows promise and key learnings about how such a device can be further optimised have been elucidated. In future work, systems could be devised such that the sensing and actuation would be completed in real-time, so that applications requiring human-like manipulation can be realised from biomedical to agricultural processes. Another unexplored research avenue is the structural health monitoring of DEAs using EIT. It may also be possible to alter the system given in this work to map the location and size of any dielectric breakdown using EIT concurrently on each compliant electrode. This would allow for more technology to be developed around the self-healing of DEAs.

6.4 CONCLUSIONS

This work set out to complete force mapping on the compliant electrode of a DEA using EIT and determine any limitations in the process. Three thicknesses of DEA with attached electrodes successfully showed actuation behaviour. Effective bulk moduli values were found to quantify the mechanical actuation impedance of each compliant electrode thickness used, ranging from 334 to 966 kPa. Force mapping was successful with decreasing degrees of mapping error with increasing compliant DEA electrode thickness. A mean centre-of-mass error of 7.9 ± 0.73 was found for the thickest used, 2 mm, compliant DEA electrode. The next steps for this research are to model the thick compliant DEA electrodes, complete a repeatability analysis of the pressure mapping performance metrics, and apply an inverse model to obtain reliable force estimates.

Chapter 7

Hardware for a DEA-EIT Sensor Actuator Hybrid Device:

7.1 Introduction

Chapter 8

Modelling of DEA-EIT Capacitively driven Hybird Sensing and Actuation Device

8.1 Objective One Conclusions

Appendix A

An Appendix

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