Screen printed textile based wearable biopotential monitoring

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Abstract

This thesis describes the development of printed wearable electrode networks on textiles for monitoring human biopotentials from the skin surface. The aim was to fabricate garments to monitor human biopotentials, such as an electrocardiogram (ECG), on a long term basis. A literature review was carried out to examine fabrication methods for wearable electrode networks on textile and screen printing is selected for this work. Several conductive and insulating screen printable pastes were then evaluated for this application and suitable pastes were selected. Screen printing was used to create networks of conductive tracks on the surface of woven textiles. These networks connect electrodes at different sites to electronics at a central location. The conductive tracks are composed of a silver polymer layer with thickness 5-10µm entirely encapsulated in polyurethane. The durability of these printed conductive tracks is investigated with cyclic stress and washing machine tests. A significant improvement in the durability of these tracks is achieved by using two different polyurethane pastes and optimising the screen printed layer structure. Tracks that can reliably endure 10 typical domestic machine washes without breaking are demonstrated. Carbon loaded silicone rubber is stencil printed to form electrodes on exposed conductive pads at the terminations of screen printed conductive tracks. The carbon loaded rubber formulation is optimised to provide electrodes with low resistivity, low surface energy and high flexibility. By using stencil printing rather than screen printing, the thickness of the electrodes is increased, causing them to protrude from the textile surface, which is useful in ensuring stable electrode-skin contact. Passive and active electrodes are fabricated on woven textiles using screen and stencil printing, and their performance is evaluated. The passive electrodes have issues with DC instability, but have suitable performance for some electromyography tasks and basic heart rate monitoring. The active electrodes show comparable performance with the gold standard, commercial Ag/AgCl electrodes. The printed textile electrode networks are demonstrated in four applications: a one-lead bipolar heart monitoring belt, a Frank configuration vector-cardiogram monitoring vest, a headband as an electromyographic (EMG) and electrooculographic (EOG) computer interface, and an armband used to examine electromyographic activity in the upper arm. Screen printing on textiles is shown to be a low-cost alternative fabrication process for durable wearable electrode networks on textiles, capable of providing high signal quality. These printed textile electrode networks are shown to be applicable to ambulatory monitoring, to reduce the associated cost and discomfort, and in hospitals and research to reduce electrode setup time.
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Glossary

**Ag/AgCl** – Refers to an electrode that uses an applied layer of silver chloride on a silver base as the electrode surface, taking advantage of the stable potential of these materials. An Ag/AgCl electrode has a stable standard potential of approximately 0.22 V.

**Baseline drift** – Baseline drift describes the signal moving from the baseline, which is 2.5 V with a 0-5 V amplifier, due to DC drift at the electrodes. Ideally the signal should have a steady baseline so that any deflection can be identified with confidence.

**Biopotential** – Any electrical potential originating in living tissue is a biopotential.

**Deposit** – In screen printing a deposit is the material pushed through the screen on to the substrate in a single print pass.

**Distal** – An anatomical term of location used to describe a point or area on a limb that is distant from the torso.

**Elasta** – Elasta is a highly elastic woven textile that is supplied in rolls of various widths. The yarns in this material are elastane. A summary of this textile’s properties is given in appendix A.

**Elasticity** – The extent to which a material returns to its original shape after being deformed.

**Escalade** – An inelastic woven textile supplied by Klopman International. The yarns are composed of 46% cotton, 38% elastane and 16% polyester. A summary of this textile’s properties is given in appendix A.

**Evolon** – Evolon is a non-woven textile produced by Freudenberg. It is composed of polyester and polyamide filaments that are entangled together with water jets.

**Flexible** – A flexible material has a low Young’s modulus.

**Inferior** – An anatomical term of location meaning “towards the feet”.

**Lagonda** – An inelastic woven textile supplied by Klopman International. The yarns are composed of 40% cotton, 30% elastane and 30% polyester. A summary of this textile’s properties is given in appendix A.

**Layer** – In this thesis a layer refers to a level of a structure composed of a single material. A layer can be composed of one or multiple prints.

**Motion artefact** – Motion artefacts are deflections that appear on an amplified biopotential signal caused by motion and pressure changes rather than the electrical activity of the organ or muscle being monitored.

**Nonwoven** – A non-woven textile is a structure composed of fibres that are combined by a process other than weaving, such as hydro-entangling. These are commonly used in flexible electronics research because their surface is much smoother than that of woven textiles.
Normal to the plane – Bending normal to the plane refers to bending where a plane is bent in to two distinct planes that are connected along the bend. For example, a 180° bend normal to the plane occurs when a sheet of paper is folded in half.

Op-Amp – An op-amp, or operational amplifier, is an electronic component that outputs the difference between two inputs. These can be used for filtering and amplification of signals.

Paste – A paste is a material with appropriate viscosity (10-30 Pa.s) and rheology (pseudoplastic) to be deposited with screen or stencil printing techniques.

Pilosity – Strands of textile yarns come loose from the weave and stick out of the textile, increasing the roughness of the textile surface. This effect is known as pilosity and makes printing more difficult.

Print – In this thesis a print refers to the material that is deposited on a structure before a curing process. A print can be composed of one or more deposits.

Proximal – An anatomical term of location used to describe a point or area on a limb that is close to the torso.

Self-adhesive – Refers here to a hydrogel that adheres to the body without requiring additional adhesive to be added. These hydrogels are used in commercial disposable electrodes.

Stiff – A stiff material has a high Young’s modulus.

Stretchability – The stretchability of a material refers to the extent to which a sheet of that material can be extended in directions parallel to the plane. A textile can have different levels of stretchability in different directions.

Skeletal muscle – A muscle used for body movement that is controlled voluntarily and usually connected to bones via tendons.

Smart textile – A textile with integrated sensing or actuating capability.

Textile – A textile is any material that is composed of a woven or non-woven mesh of small fibres.

Via – A via is a conductive path through a sheet of material, electrically connecting the two surfaces. In this thesis it is mainly used to describe a conductive path connecting the front and back of a textile.

Vulcanisation – The process of crosslinking polymer chains in a liquid rubber to form a solid rubber.

Warp – Yarns that are held in tension during the weaving process, so the textile typically has less stretchability in this direction. These yarns are at 90° to the weft.

Weft – Yarns woven between the warp yarns, so the textile typically has more stretchability in this direction. These yarns are at 90° to the warp.

Young’s modulus – This is a measure of the flexibility of a material. It is calculated by dividing the stress applied to a material by the extension of the material due to that stress.
List of Acronyms

**ADC** – Analogue to digital converter

**DI** – Deionised

**ECG** - Electrocardiography

**EEG** - Electroencephalography

**EMG** - Electromyography

**EOG** – Electrooculography

**FES** – Functional electrical stimulation

**PCB** – Printed circuit board

**PDMS** – Polydimethylsiloxane

**phr** – Per hundred rubber (lower case to conform to its use in the literature)

**RMS** – Root mean square

**SEM** – Scanning electron microscope

**SD** – Standard deviation

**UTCP** – Ultra-thin chip packaging

**VCG** – Vectorcardiogram
1 – Introduction

This thesis examines screen printing as a fabrication method for wearable biopotential monitoring systems. A wearable biopotential monitoring system comprises four main components, which are described in Table 1-1.

<table>
<thead>
<tr>
<th>Component</th>
<th>Function</th>
<th>Location</th>
</tr>
</thead>
<tbody>
<tr>
<td>Skin surface electrodes</td>
<td>Adhere to the skin and conducts biopotentials.</td>
<td>Sites of interest.</td>
</tr>
<tr>
<td>Electronics</td>
<td>Record and store signals from sensors.</td>
<td>Anywhere that will minimise discomfort.</td>
</tr>
<tr>
<td>Conductive paths</td>
<td>Electrical connections between the sensors and the electronics.</td>
<td>Between the sensors and electronics.</td>
</tr>
<tr>
<td>Textile garment</td>
<td>Makes the system wearable.</td>
<td>Substrate for the other components.</td>
</tr>
</tbody>
</table>

In many cases the system also has a wireless capability for the transmission of signals to allow remote monitoring. As this thesis investigates the fabrication of such systems by screen printing, the main focus is on electrodes and conductive paths. These have various requirements as described in Table 1-2.

<table>
<thead>
<tr>
<th>Electrodes</th>
<th>Conductive paths</th>
</tr>
</thead>
<tbody>
<tr>
<td>Robust coupling to the skin</td>
<td>Biocompatible</td>
</tr>
<tr>
<td>Appropriate electrical properties</td>
<td>Waterproof</td>
</tr>
<tr>
<td>Biocompatible</td>
<td>Durable</td>
</tr>
<tr>
<td>Waterproof</td>
<td>Flexible</td>
</tr>
<tr>
<td>Durable</td>
<td></td>
</tr>
<tr>
<td>Comfortable</td>
<td></td>
</tr>
</tbody>
</table>

There have been several projects aimed at introducing a viable proposal for wearable biopotential monitoring. Major prototypes from the last decade include VTAMN [1], MyHeart [2], iTCares [3], MagIC [4], WEALTHY [5], Smartshirt [6] and Lifeshirt [7]. The European Commission committed 60 million euros to the research and development of ‘personal health systems’ under framework package 7 [8]. These systems all use conductive yarns to implement electrodes and interconnections. The concept of using printed systems for this purpose has been relatively unexplored.

This chapter introduces the objectives and motivation for printed wearable biopotential monitoring systems, novelties and publications arising from this work, and the structure and content of this thesis.
1.1 Objectives of this work

This work aims to find a printed textile solution for biopotential monitoring with a primary focus on ECG monitoring. The aim is to use screen printing to create a garment with skin contact electrodes and conductive paths that are durable under conditions such as mechanical stress and moisture during both the wearing and washing of the system. The performance of the electrodes should be optimised and the fabrication process should be homogenised. The incorporation of the electronics is considered but a portable electronics module is outside the scope of this project.

This work was partially funded by the European Commission under Framework package 7 as part of the BRAVEHEALTH project (Patient Centric Approach for an Integrated, Adaptive, Context Aware Remote Diagnosis and Management of Cardiovascular Diseases). The goal of this project was to fabricate a wearable vectorcardiogram (VCG) monitoring garment with electrodes in the Frank configuration.

1.2 Motivation for this work

Wearable health monitoring systems are required because of a shift in the application of medicine from treatment to prevention. A health service can work more efficiently by avoiding illness pre-emptively rather than treating each case once it becomes a hindrance to the individual [9]. There is also a significant argument in favour of wearable health monitoring as a clinician can utilise more information about a patient recorded over a longer period than would be available in a single clinical session, which theoretically allows a more accurate diagnosis.

Remote monitoring allows greater freedom of movement, as the system is implemented to be as unobtrusive as possible. This means that patients do not have to remain in hospital, reducing costs and freeing up resources for other patients. The networks of wireless and wired information channels in developed nations make remote monitoring possible; Varshney et al [10] state that “with an increasingly mobile society and the worldwide deployment of mobile and wireless networks, the wireless infrastructure can support many current and emerging healthcare applications.”

A washable garment for ECG monitoring could have several applications. They are already used in athletics and fitness. 12-lead examinations in a clinical setting could benefit from a single garment to reduce the setup time and cost of disposable electrodes. Wearable systems would make possible the recording of ambulatory morphological ECG data with high density of electrode networks and predefined electrode locations on the body. A reusable system would also be more cost-effective in this setting than existing Holter monitors as the electrodes do not have to be replaced.

Screen printing is a novel fabrication method for wearable biopotential monitoring. It allows more freedom of design to create customised three-dimensional sensor structures compared to existing methods using conductive yarn. Printed tracks are easier to cross over without shorting and easier to connect to external devices. It is possible to create dense networks of conductive paths with printing that are unfeasible with conductive yarn.
1.3 Statement of novelty

The novelty in this work originates from the fabrication method. Printing is already a common procedure in the textile industry which allows more design freedom in wearable monitoring systems than existing methods using conductive yarn. Although examples of individual printed wearable sensors exist in the literature, a fully integrated system on a woven textile substrate in which both sensors and encapsulated interconnections are printed is realised for the first time in this thesis. The specific novelty claims in this work are given here.

- A method of screen printing electrodes and connections on textiles for bio-potential monitoring.
- An optimised formulation of carbon loaded silicone rubber for durable, dry electrodes.
- A detailed analysis of the suitability of several UV curable polyurethane pastes as an interface material for conductive tracks on textile.
- An analysis of carbon-loaded silicone rubber electrodes and a comparison with conventional disposable electrodes.
- Screen and stencil printed active electrodes on woven textiles.
- A screen-printed-textile active electrode wearable chest band for bipolar one-lead holter monitoring.
- A screen-printed-textile Frank configuration vectorcardiogram monitoring vest.
- A screen-printed-textile headband for computer control using facial electrooculography and electromyography.

1.4 Publications arising from this work

Seven publications have arisen from this work. Five of these are already published, with two further publications awaiting peer review. The published articles are a poster session, a conference paper and three journal papers. The publications are listed here in chronological order.

1.5 Structure and content of this thesis

This thesis is split into ten chapters which address the context, fabrication, optimisation and evaluation of textile-based screen and stencil printed networks for biopotential monitoring.

Chapters 1 and 2 cover the context of this research. Chapter 1 has described the objectives and motivation for this research and described several novelties arising from this work. Chapter 2 puts these objectives and novelties in a wider context with an extensive literature review covering biopotential monitoring, electrode design and smart textile research.

Chapters 3, 4 and 5 cover the fabrication of the devices described in this thesis. Chapter 3 gives design requirements and discusses how the properties of textiles affect the design of devices. The overall layout and the designs for individual components such as electrodes and vias are described. Chapter 4 provides information on the selection and formulation of printable pastes that are used to fabricate the structure. Chapter 5 gives details about the fabrication process using the selected pastes and textiles.

Chapters 6, 7 and 8 discuss the performance and optimisation of the developed electrode networks. Chapter 6 describes the optimisation of the durability of printed conductive tracks using composite polyurethane structures. Chapter 7 examines the performance of the developed printed passive electrodes and printed active electrodes. Chapter 8 describes several electrode networks designed for specific applications and evaluates their performance in these applications.

Chapter 9 concludes this thesis with a summary of work, a discussion and recommendations for future work.

This thesis includes a glossary and list of acronyms before the introduction. A reference guide to the properties of pastes and textiles, standard test apparatuses and screen designs used in this thesis can be found in the appendices.
2 – Literature review

2.1 Introduction

This literature review aims to provide a background regarding the biopotential monitoring and screen printing technology examined in this thesis. This review examines the theory behind this technology as well as looking at examples of the state of the art in wearable biopotential monitoring and smart textiles. The four sections here discuss human biopotentials, wearable health monitoring, smart textile fabrication techniques and screen printed smart textiles.

2.2 Human Biopotentials

This first part of the literature review gives a background on human biopotentials. Human biopotentials on a cellular level can be measured with needle electrodes, but those measured from electrodes on the skin surface originate in muscles or in the brain and provide indicators as to the behaviour of large groups of cells. The first section will discuss three commonly researched biopotential types. The practicalities of monitoring biopotentials, such as signal acquisition and electrodes, are then discussed in the second section.

2.2.1 Types of human biopotential

This section will discuss electromyography, a general term for electrical potentials originating in muscles, electrocardiography, which describes electrical potentials that originate in the heart, and electroencephalography, which describes electrical potentials that originate in the brain.

2.2.1.1 Electromyography (EMG)

Electromyography is a term that describes the monitoring of electrical signals produced by skeletal muscles. Although electrocardiography is the main focus of this research, electromyography is described first in this thesis to provide a basic understanding of the electrical behaviour of muscle cells.

Muscle cells (myocytes) are triggered as a group by an electrical signal from a motor neuron. The group of myocytes that function based on the signal from a single motor neuron is known as a motor unit. A muscle can contain one or more motor units. When a motor unit is triggered the myocytes depolarise and shrink, and the muscle contracts. This also causes a change in the potential at the skin surface.

The monitoring of electrical signals originating in the arm, leg and facial muscles are the most common applications for EMG. Any muscle that contracts produces an electrical potential that can be observed, however some are difficult to observe from electrodes on the skin surface, and instead require invasive EMG with needle electrodes.

Both AC and DC potentials can be observed in EMG measurements. When a muscle is contracted repeated motor neuron firing is required to maintain the same level of contraction. This repeated firing and the response of the cells in the muscle (myocytes) can be observed as
an AC signal on the skin surface. These AC potentials tend to have RMS amplitudes between 0.01-1 mV on the skin surface. They are typically amplified with gains of 1000-10000 to be read.

This AC EMG effect is shown in work carried out by Van Boxtel [11] in which the electrical behaviour of the corrugator supercili, the facial muscle that pulls the eyebrows down, is observed. This muscle is a single motor unit that can be controlled voluntarily, so the motor impulse train is easy to extract. The nerve impulses can be observed in the surface EMG readings. The surface EMG and the extracted impulse train are shown in Figure 2.2-1.

![Surface EMG signal and extracted motor unit impulse train from work by Van Boxtel [11].](image)

DC potentials, by comparison, are observed between two electrodes when one is monitoring a contracted muscle and the other is not, with the depolarised myocytes causing a DC potential difference on the skin surface. These DC potentials provide useful information when two muscles are normally used in such a way that if one is contracted the other is not, such as in the control of the eye or the wrist.

### 2.2.1.2 Electrocardiography (ECG)

ECG is the most commonly examined parameter in wearable health monitoring. One of the reasons it is important is the prevalence of heart problems in the general population. A report from the American Heart Association states that ‘Coronary heart disease caused approximately 1 of every 6 deaths in the United States in 2006’ [12]. This means that preventative solutions are desirable, both for the lives they can save and for the reduced financial cost of prevention as compared with treatment.

Several tools are used diagnostically by the cardiologist, including chest X-Rays, radiolabelled isotopes inserted into the blood stream (nuclear cardiology) and the insertion of small tubes into the heart (catheterization). The electrocardiogram is, however, the most important diagnostic tool, because it is safer and faster than the other techniques mentioned. Dymond and Broadhurst [13] state that ‘the ECG is an essential investigation in the assessment of the cardiac patient.’

The heart is made up of two types of myocytes. Cardiomyocytes make up the left and right atria and ventricles, and are responsible for the pumping action of the heart. Cardiac pacemaker cells are responsible for the excitation of the different groups of cardiomyocytes at
the appropriate time, based on signals from the brain. A diagram of the main electrical conduction elements in the heart is shown in Figure 2.2-2.

![Diagram of the heart](image)

**Figure 2.2-2:** A diagram of the main electrical conduction elements in the heart. [14]

In each heart contraction electrical excitation begins in the sino-atrial (SA) node in the right atrium and spreads to the atrio-ventricular (AV) node and then on to the ventricles. This electrical excitation triggers the muscular contraction of the heart that pumps blood around the vascular system. Thousands of cells doing this together form a wave of polarisation or depolarisation. This creates relative electrical potential differences on the skin surface with a magnitude of around 1 mV [13]. An amplifier with a gain of 1000 is normally used to amplify these signals up to easily readable levels.

As with all voltages the cardiac potentials have direction as well as magnitude, so switching the amplifier inputs will invert the observed potential. Because of this a more detailed ECG diagnosis can be made by examining the potentials from several angles; this is why seven or ten electrode setups are often used in diagnosis. The same deflections are observable from most angles except in the case where the potential is at an angle of 90° to the electrode pair. The magnitudes of these deflections, however, are different and from some angles inverted. The deflections, as shown in Figure 2.2-3, are denoted with the letters P, Q, R, S, T and U.

![ECG signal](image)

**Figure 2.2-3:** A simulated skin-surface ECG signal. [15]
Each deflection corresponds to a particular activity in the heart. The medical explanation of each deflection is beyond the scope of this thesis, however the P-wave is discussed to give an example of how the smaller deflections (P, T and U) can be diagnostically useful.

The P-wave corresponds to the initial depolarisation of myocytes in the SA node. The mass of muscle in the atria is relatively small and the size of the P-wave corresponds with this. An abnormal P-wave could show that the atrium is an abnormal size or that depolarisation originates elsewhere. There is an interval between the P wave and the QRS complex. If this interval is too long it could show that conduction from the SA node to the ventricular system is in some way inhibited. This illustrates the importance of a wearable ECG detecting the small as well as the large ECG deflections.

One of the goals of this work is to produce a printed Frank configuration vectorcardiographic garment. This uses seven electrodes arranged through a resistive network into three bipolar channels and a ground electrode to give a three dimensional view of electrical activity in the heart. The placement of the differential electrodes is crucial, as the heart must be viewed by three separate channels in the X, Y and Z directions. Unlike the 12-lead configuration, which uses six electrodes across the heart and one on each limb, most of the electrodes in the Frank configuration are on the torso so it is appropriate for integration into a single vest garment with modified positions on the head (H) and foot (F) electrodes. The electrode positions and resistive network are shown in Figure 2.2-4.

![Figure 2.2-4: Frank configuration electrode positions and resistive network. [16]](image)

2.2.1.3 Electroencephalography (EEG)

EEG involves monitoring electrical signals produced by the brain. This is especially useful in monitoring sleep disorders and conditions such as epilepsy, however there is also research into
monitoring mood, alertness, activity and intent through processing electroencephalographic signals.

There are two main types of brain electrical activity that can be observed from the surface. One is repetitive waves of various frequencies, which can be used as indicators for the level of alertness or concentration in an individual. The other is the event-related potential (ERP) which can be observed after a sensed event, usually a light or noise. EEG signals typically have amplitudes around 10-100 µV on the skin surface. Because of this they are typically amplified with a gain of 10,000 in order to provide a signal amplified to a useful level. Since the amplification is high, noise must be minimised to acquire a reliable signal.

2.2.2 Human biopotential monitoring with skin-surface electrodes

This review now examines methods of monitoring the potentials described above with electrodes placed on the skin surface.

2.2.2.1 Signal acquisition

The signal acquisition process can vary to suit different requirements in hospital monitoring and ambulatory monitoring situations, so only the outline of the process is described here. The typical process uses the difference in potential between two ‘differential’ electrodes. These electrodes are placed on the skin and connected to the inputs of an instrumentation amplifier. The amplification stage is important in ensuring the signal can be read. An instrumentation amplifier is used because it has a very high common mode rejection ratio; any signal that is the same on both electrodes makes very little impact on the output. The gain is usually in the range 1,000-10,000.

In older systems the output of the instrumentation amplifier could be connected directly to a stylus but modern systems tend to sample the output with an ADC and display or store the information digitally. In a remote system the method is the same except that the data is periodically transmitted rather than stored. Figure 2.2-5 shows a block diagram of the signal acquisition setup.

![Diagram](image)

**Figure 2.2-5:** Typical signal acquisition setup for remote and clinical monitoring.

There are two ways in which an acquired signal can be distorted and inaccurately represent the potential at the skin surface. One is baseline drift, also referred to as signal drift on DC drift. Variations in the DC potential at the skin surface can cause a change in the base level of the signal, which can be identified incorrectly as a muscle signal. In the worst case this can cause the signal to saturate at the voltage output limits of the instrumentation amplifier and prevent...
any useful data from being collected. The causes behind such variations are examined experimentally in later chapters.

The other type is AC noise of which by far the most prevalent is power line noise, which has a frequency of 50 Hz in the UK. Capacitive coupling, where the signal couples directly into the cables or skin-electrode capacitances, is one way this occurs. The other is noise that couples into the body and is observable from the skin surface, even if the electrodes and cables are noise-less. AC noise is referred to simply as noise in this thesis.

Measurements with two electrodes, in which the two are directly compared, are called unipolar measurements. By contrast, a bipolar measurement uses three electrodes. Two of these are ‘differential’ electrodes and the third is a driven right leg (DRL) electrode. The DRL reduces the voltage difference between the ground of the amplifier and the DC voltage of the patient by effectively driving the voltage of the patient to the common mode voltage of the amplifier [17]. Consequently, there is reduced common mode voltage and reduced noise. Recordings in this thesis are bipolar measurements.

2.2.2.2 Electrode preparation

Electrodes can be used ‘wet’ or ‘dry’. This refers to the skin preparation rather than the electrode construction. ‘Wet’ means that an electrolyte paste is used to reduce the resistance and capacitance between the skin and electrode. In most clinical cases, ‘wet’ electrodes are used. The electrode paste must be re-applied for each application of the electrode. This is largely unsuitable for wearable health monitoring systems. Pacelli et al [18] note that in the MyHeart project the aim was to interface the electrodes directly with the skin, and so an electrode paste is not used.

The electrolyte paste wets the skin and consequently reduces the skin impedance. There is a trade-off in wearable electrodes between the comfort and ease of use of ‘dry’ electrodes and the low skin impedance resulting in higher signal quality with ‘wet’ electrodes.

Searle et al [19] perform a simple experiment to show the difference in noise level caused by input impedance mismatch. Three electrode pairs consisting of wet-wet, dry-dry and wet-dry combinations have noise measurements taken. The wet-dry pair has the highest RMS noise, and the wet-wet pair has the lowest.

Table 2-1: Noise level using different electrode pairs in work by Searle et al. [19]

<table>
<thead>
<tr>
<th>Electrode Pair</th>
<th>Noise Level (µV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Two Ag/AgCl (Wet)</td>
<td>3.2</td>
</tr>
<tr>
<td>Two stainless steel (Dry)</td>
<td>74.2</td>
</tr>
<tr>
<td>One Ag/AgCl, one stainless steel</td>
<td>762.7</td>
</tr>
</tbody>
</table>

Xu et al [20] describe another issue; the ‘wet’ electrodes will gradually dehydrate causing the signal to deteriorate over time. However, factors such as perspiration can also greatly affect the conductivity in either type of electrode. In this thesis electrode pastes and gels are not used as ‘dry’ electrodes require no preparation and are stable over longer periods.
2.2.2.3 Types of electrodes

This section describes the different types of electrodes used for skin surface biopotential acquisition.

**Hydrogel and sponge electrodes:**

Hydrogel and sponge covered electrodes are popular in biopotential measurements because they allow ionic flow between the electrode and the skin. The electrode is usually silver with a silver chloride layer, commonly referred to as an Ag/AgCl electrode. An electrode double layer is formed between the ions in sweat or electrode paste that pass through the hydrogel and the Cl\(^{-}\) ions from the electrode. This electrode potential is very stable in Ag/AgCl electrodes, which prevents changes in DC potential at each electrode, minimising the drift of the observed signal. When this is a path for ion conductivity between the skin and the silver chloride layer the effective resistance of the electrode is very low.

A NASA silicone-sponge electrode from a technical brief published in 1972 [21] is shown in Figure 2.2-6. The top of the electrode is cut off and the electrode is placed on the skin, allowing ionic flow between the silver/silver chloride disc and the skin surface. Dimensions are not provided with the design.

![Figure 2.2-6: A silicone-sponge electrode sealed in a vinyl coating for use as a disposable electrode for monitoring astronauts [21].](image)

Hydrogels and sponges can also reduce motion artefacts, as they allow small movements and deformations of the electrode without sliding across the skin surface [22]. The disadvantage of these types of electrode is the fact that the gels can provide a favourable environment for bacterial growth, meaning the risk of infection can be high. These electrodes can be self-adhesive, in which the hydrogel itself is tacky and makes the electrode adhere to the skin, or have an adhesive in a ring around the electrode. In either case, hairs and dirt can gather on the adhesive part, reducing the lifetime of the electrode. Hydrogel and sponge electrodes are not developed in this thesis as their durability is low and they are not reusable.
Wet electrodes:

Wet electrodes provide a disposable solution for biopotential monitoring. They are composed of an Ag/AgCl electrode that is already covered with an electrolyte paste. These provide lower electrode resistances than hydrogel and sponge electrodes without paste because they hydrate the skin instantly and have no solid barrier between the skin and the Ag/AgCl layer. Although hydrogels and sponges are porous they are solids in nature and therefore reduce ionic flow. These electrodes rarely require an additional electrolyte paste and are an ideal solution for fast, high quality recordings. However, they are not durable and have problems with shorting between electrodes if the inter-electrode distance is small. Further, these electrodes dry out over time, greatly reducing the signal quality [19]. Consequently this type of electrode is not suitable for long term monitoring.

Textile electrodes:

Textile electrodes are electrodes constructed by weaving, knitting or embroidering conductive yarns to fabricate an individual electrode. Knitted electrodes are standalone components while embroidered electrodes are fabricated directly onto a textile garment. This technology is very durable and has been demonstrated in the acquisition of ECG in several situations. A knitted stainless steel electrode with a skin contact area of 3 x 3 cm fabricated in work by Catrysse et al [23] is shown in Figure 2.2-7 as an example.

![Figure 2.2-7: A 3 x 3 cm knitted stainless steel textile electrode [23]](image)

There are, however, several issues with this technology. Although it integrates very well with textiles, it is difficult to connect conductive yarns to electronics in durable manner. This means difficulty not only with connecting electrodes to a central amplifier, but also means it would be very difficult to integrate electronic components to fabricate an active textile electrode. It would also be difficult to create a textile electrode that protrudes from the textile surface without using an excessive quantity of yarn or using an additional fabrication technique.

Textile electrodes are also difficult to use to create an electrode double layer as in silver/silver chloride hydrogel electrodes. Coosemans et al [24] noted that baseline drift was significant on their passive textile electrodes, as there was no stable electrode double layer. Researchers have successfully applied silver chloride to silver conductive yarns to create Ag/AgCl yarn electrodes [25] by electroplating in a chloride solution, however these electrodes must have a
chloride layer reapplied after use. ECG readouts from passive textile and silver/silver chloride electrodes are shown in Figure 2.2-8.

![ECG Readouts](image)

**Figure 2.2-8:** (a) ECG obtained with textile electrodes, and (b) ECG obtained with silver/silver chloride electrodes. The same amplifier was used for both recordings. [24]

**Conductive rubber electrodes:**

A dry conductive rubber ECG electrode has been previously achieved by several authors in the literature. Linti et al [26] fabricated a system to monitor the physiological parameters of infants. In this system conductive rubber electrodes were used to monitor the ECG. These electrodes use silicone rubber loaded with silver particles. They are connected using several polytetrafluoroethylene-insulated 36-AWG wires that are woven into a textile ribbon. The electrodes are durable and waterproof, however the wires must be picked individually from the textile ribbon and connected to the sensors by hand. An SEM cross-section of an electrode is shown in Figure 2.2-9 and the completed system is shown in Figure 2.2-10.

![SEM Cross-Section and Completed System](images)

**Figure 2.2-9:** An SEM cross-section of a silver loaded silicone rubber electrode on textile [26].

**Figure 2.2-10:** The completed system designed and fabricated by Linti et al [26].

Conductive rubber is preferred to a hydrogel or sponge because of its low surface energy, which prevents hairs and other dirt from attaching to the electrode. The author’s practical experience has shown that this limits the life of hydrogel electrodes. There will also be less chance of skin irritation and bacterial growth. The lack of silver/silver chloride/ionic fluid mechanism with conductive rubber means that baseline drift with conductive rubber
electrodes is greater than with Ag/AgCl electrodes, as conductive rubber has a much less stable electrochemical potential.

**Active electrodes:**

Active electrodes reduce the interference in an electrode signal by inserting a buffer amplifier at the site of the electrode. The buffer amplifier has unity gain and so does not amplify the signal, but it reduces the effective impedance of the signal as seen by the instrumentation amplifier. This means that there is very little impedance mismatch between differential electrodes and consequently low electromagnetic interference compared with passive electrodes. The input impedance of the buffer amplifier is very high, so the electrode’s output signal is affected less by high impedance sources, reducing the need for any skin preparation.

Information from recent publications shows that active electrodes perform comparably to even very high quality commercial passive electrodes in all circumstances [19], [27]. They do, however, require electronic components at the site of the electrode, as well as positive and negative power supplies for the op-amp. This thesis focuses on the materials, durability of structures and proof of concept for screen printed textile-based biopotential monitoring. However, if dry electrodes are implemented in these designs, fabricating active dry electrodes would improve the signal quality to match that of existing disposable electrodes. An active electrode circuit from work by Kang et al [27] is shown in Figure 2.2-11, while Figure 2.2-12 shows a comparison between active and passive electrodes and the corresponding contact impedance from work by Searle et al [19].

![Figure 2.2-11: Active electrode structure used by Kang et al [27]. R = 200 kΩ, R = 50 kΩ, and C = 10 nF.](image)

![Figure 2.2-12: Comparison of dry passive and active electrodes in work by Searle et al [19].](image)

**Capacitive electrodes:**

Capacitive electrodes do not form a conductive path between the skin surface and the amplifier. Instead, they form a capacitor with the skin as one plate and the electrode as the other. The most obvious advantage of this is that the capacitor does not permit any DC flow and there is no half-cell potential formed at the electrode surface. This greatly reduces the effect of motion artefacts. There is also no requirement for skin preparation.

The issue with capacitive electrodes is that the signal is weak, consequently the signal to noise ratio is low and these electrodes require on-electrode amplification and shielding. In work by Chi et al [28] a set of capacitive electrodes were produced that gave high quality ECG readings.
However, each electrode required two onboard PCBs and a shielding enclosure. The electrodes are shown in Figure 2.2-13 and ECG readings through a ‘thin T-shirt’ and through a ‘thick sweater’ are shown in Figure 2.2-14.

Given the required level of complexity for this type of electrode it was felt that attempting to create such an electrode by printing on textiles would be beyond the scope of this research.

2.2.2.4 Sources of signal noise

Power line noise:

The main source of noise in any biopotential monitoring system is power line noise, or mains power noise. This is sometimes referred to by the corresponding frequency, 60 Hz in the United States and 50 Hz in the UK and Europe. It is generally accepted that there will be some 50 Hz noise, however when it is equal on both electrodes it is rejected because of the high common mode rejection ratio of an instrumentation amplifier. There are two mechanisms by which this can couple into the system.

The first coupling mechanism is electrode impedance mismatch. If the impedances of two differential electrodes are mismatched, the 50 Hz noise will appear larger at one terminal than the other, even though the magnitudes of the sources are the same. This is because the impedance of the signals is very high and therefore the current and power of the signal is very low. When a signal is very low power the operational amplifiers will pass a lower voltage than that which is actually at the input; common mode signals will have different magnitudes if the electrodes have different impedances. By minimising skin-electrode impedance on all electrodes, the noise on a signal is minimised. The skin-electrode impedance is comprised not only of the electrode but also the impedance between the skin and electrode, and the impedance of the skin itself [19].

The other significant source of noise is capacitive coupling, where noise couples capacitively into the cables connecting the electrodes to the amplifier. Shielding cables or twisting cable pairs together to ensure any noise is common mode are common methods of reducing interference coupling into the cables. The capacitive coupling mechanism means that
capacitive electrodes, where there is no skin contact, have a very high sensitivity to electromagnetic noise.

**Motion artefacts:**

Movement of an electrode relative to the skin causes motion artefact. There are several ways that this effect can manifest. A change in pressure on an electrode causes a DC drift in the signal, as the pressure affects the DC potential seen at the skin relative to another differential electrode at stable pressure. Pressure changes are of significant interest in a textile system, where the electrodes are held to the skin by a garment rather than by an adhesive, and pressure changes are likely to be more significant.

If a motion lifts part of an electrode from the skin surface the impedance of that electrode can be altered and the result is increased noise and DC drift. This is a common occurrence during the use of lower quality self-adhesive electrodes which results in increased noise due to impedance mismatch.

**Biopotential cross-sensitivity:**

Voluntary muscle contractions, especially in the upper arm or the chest, cause electromyographic signals to be picked up in the ECG. There is no way of accounting for this in electrode design, this noise can only be minimised by electrode positioning and signal processing.

**Other sources of noise:**

Any technology that gives off electromagnetic radiation could conceivably be picked up on a biopotential measurement, given sufficient amplitude and proximity. Fluorescent bulbs are cited as one such source of noise. Most electromagnetic radiation is relatively high frequency, and is unlikely to have a great influence at the typical sample rates of 500 Hz or 1000 Hz used in ECG, EMG and EEG.

Noise in the equipment can also influence signals. There will be a very small amount of noise in the amplifier, which can usually be neglected. The digitisation of the ADC can also have an influence. In this thesis a USB-6008 data acquisition device is used as an ADC. The maximum resolution is 10 mV in the range 0-5 V, which means that it is impossible to distinguish different levels of noise in signals which have peak to peak noise amplitudes of under 10 mV. The datasheet for the device also states that the RMS noise is around 3 mV.

**2.2.3 Conclusions**

The work is primarily aimed at fabricating textile based signal acquisition hardware for heart monitoring. However, given the similarity of the acquisition technique for other human biopotentials, it is useful to explore the versatility of the fabricated hardware by examining its suitability for other muscle monitoring applications. EEG will not be examined, however, because the electrode technology is in the early stages of development and the low signal amplitudes of EEG require a high signal to noise ratio. EMG contractions are easier for testing
because the movement of a skeletal muscle can be voluntarily triggered and the signal amplitude is greater.

It is necessary for dry electrodes to be designed in this work, given the need for durability and stability over long periods of time. Hydrogel, sponge and wet electrodes gradually deteriorate and have to be replaced or re-moisturised. Dry electrodes have problems with high impedance, so active electrodes should be investigated in order to overcome this difficulty and provide better signal quality.

2.3 Wearable monitoring systems

This section discusses the state of the art in wearable systems to monitor physiological parameters, with a focus on systems for biopotential monitoring.

2.3.1 History of wearable monitoring

Wearable monitoring systems fabricated on textiles are a relatively new concept in medicine, although Holter ECG monitoring with electrodes and a wireless transmitter was first demonstrated in 1949 [29]. Other examples of early wearable electrode networks can be seen in EEG and EOG monitoring of astronauts in the 1970s [21].

Georgia Institute of Technology is credited with having carried out the earliest research resulting in a textile based wearable monitoring system beginning in 1998 [30]. Their system was proposed in response to a US Navy call for papers and a project proposal for a system to monitor soldiers’ vital signs and alert the medical triage to the penetration of a bullet. This was implemented first with fibre optic cables and then with woven and knitted conductive yarn and is shown in Figure 2.3-1.

![Figure 2.3-1: The Georgia Tech Wearable Motherboard [30].](image)

In 2002 printing was considered for the same purpose in research by North Carolina State University [31]. They began research by attempting to screen print coplanar waveguides, a two-dimensional form of shielded cable, on to nonwoven textiles. They also examined the durability of this printed structure when washed and developed screen printed electrodes on non-wovens. This work is examined in more detail in Section 2.5.2. Printed silver coplanar waveguides on Resolution Print Media, a non-woven textile, are shown in Figure 2.3-2.
2.3.2 Components of a wearable monitoring system

Wearable monitoring hardware can be split into four parts; sensors, conductive paths, a garment and electronics. This thesis is focussed on the wearable system, composed of electrodes (as the sensors), conductive paths and a garment. Although some consideration is given to methods of amplification and electronics integration, the electronics are not examined in detail. This section discusses these four components. A concept diagram of a typical wearable monitoring system with two types of sensor, such as ECG electrodes and temperature sensors, is shown in Figure 2.3-3.

![Figure 2.3-3: A concept diagram of a typical wearable monitoring system with two types of sensor.](Image)

2.3.2.1 Sensors

The sensors integrated into wearable monitoring systems are often electrodes and strain sensors but other sensors such as accelerometers [4] and temperature sensors [32] have been integrated into wearable electronics. The design of the sensor elements as well as their placement on a garment can have a significant effect on the performance of the sensor.
With a textile skin-contact biopotential electrode, poor placement on the body could result in loss of electrode contact during movement. The placement of the electrodes relative to the electronics can also affect performance. The conductive paths in a biopotential sensing garment are often not shielded due to fabrication limitations. Consequently, increasing the distance from the electrodes to the electronics increases the noise observed on the signal.

2.3.2.2 Conductive paths

These connect the sensors to the electronics. In most cases, the low resistivity found in standard cabling cannot be achieved with conductive yarns or printed tracks. With both fabrication techniques, the requirement for flexibility dictates that the conductive material cannot be pure metal. In printed tracks the conductive ink or paste is usually composed of particles of conductive material in a polymer binder, while conductive yarns are composed of a combination of conductive and structural fibres.

Conductive paths must introduce as little noise as possible into the system. In wearable systems the cables often cannot be fully shielded for fabrication reasons. For this reason, some biopotential monitoring systems place the electrodes near the amplifiers or use standard shielded cables so that interconnections via a textile substrate are either very short or not required. It should be noted, however, that in clinical ECG exams with wet or hydrogel Ag/AgCl electrodes the cables are normally not shielded as the electrical impedance of the electrode-skin interface is very low, and consequently the signals are relatively low-impedance and less susceptible to noise.

In biopotential measurements increased connection resistances up to 100 Ω are usually acceptable, as the skin-electrode resistances are typically on the order of kΩ with Ag/AgCl electrodes and hundreds of kΩ with dry rubber electrodes. These low conductive path resistance variations will affect the resulting signal only negligibly. The resistance of the conductive path should not increase to the region of kΩ with factors like strain, temperature or humidity, as this will be detrimental to the signal. Both printed tracks and conductive yarn paths are prone to significant changes in resistance or even a break in the conductive path after use and after washing, which render the system unusable. The design of interconnections is important as they must remain intact for the operating life of the system.

2.3.2.3 Garment

The garment must be comfortable and breathable to be worn on long term basis. In a dry electrode system the garment must also be designed to hold the electrodes to skin, as there is no adhesive used.

The garment must also be suitable for the fabrication process. In this thesis printing is used and the smoothness of the substrate surface affects the printability. The thicker the yarn in the textile is the rougher the surface will become, and rough surfaces are more difficult to print on. Many textile circuit printing research prints onto either non-woven textiles, such as Evolon or Tyvek, or onto a textile with very thin yarns such as silk. Although these substrates are easier to print on, they are less practical for a durable garment because they are not breathable.
2.3.2.4 Electronics

These frequently include a wireless communication module as well as a power supply, analogue front end amplification, ADC, a digital signal processor and memory. The electronics process the signals from the sensors and either store or transmit a readable output for diagnosis. Electronics are not examined in detail in this thesis although where they affect the design of electrodes, conductive paths or the garment they are considered.

2.3.3 Example system

MagIC is a system developed by various Italian institutions. It began production around 2003 and received the CE marking in June 2009. It has been reported in several papers as it has been further developed and tested [33] [34] [35].

The stated aim of the MagIC system is ‘the unobtrusive recording of cardiorespiratory and motion signals during spontaneous behaviour’. The information here is drawn from these sources. This aim is illustrated by some of the situations in which this system has been tested, such as on a climb of Mount Everest and during skydiving. For this purpose the system is designed to be lightweight, unobtrusive and have the capability for monitoring signals on a long term basis. One of the iterations of the MagIC system is shown in Figure 2.3-4.

Figure 2.3-4: The MagIC system as reported in [35].

In terms of fabrication, the MagIC project is very compatible with textile manufacturing technology. It also has a long battery life and a simple design. These are key features in a system designed to be unobtrusive. The system comprises an ECG, a breathing sensor and a 3-axis accelerometer. The ECG is a one-lead ECG implemented with two electrodes positioned on opposite sides of the thorax. It is stated that these electrodes are woven however the yarn used is unspecified. The breathing sensor is implemented with a conductive yarn constructed of an elastic core with a conductive yarn wrapped around it. Again, the conductive yarn used is not named. The accelerometer is integrated into the electronics module, which is positioned on the garment just above the hip. In some conditions the system can also be used to monitor the heart using a seismocardiogram, using an accelerometer to detect the pulsating deflection of the skin on the chest during a heartbeat. The vest itself is made of cotton and Lycra and is fully washable.
There are, however, limitations to this system. The electrodes are not shielded, nor do they have any on-site buffering. The signal quality is improved slightly by using large electrodes and minimising the distance to the electronics, but this places limitations on design freedom. As the ECG signal quality is low this system could only be used for detecting arrhythmia, and is not reliable regarding morphological data. This also applies to the data from the seismocardiogram. The connection method used to connect conductive yarns to the electronics is not described.

2.3.4 Conclusions

Several systems for the monitoring of physiological parameters exist, with the ECG being the most commonly measured parameter. Lists of wearable monitoring systems in development currently or through the past decade can be found in reviews by Cho et al [36] and Xu et al [20]. These reviews show comparatively little information on printed systems.

Where standard clinical electrodes are not used, conductive yarns are usually employed for the production of wearable electrodes and connections. These have the advantages of being durable and textile compatible. This technology, however, limits the thickness of electrodes, meaning any garment must be close fitting to ensure stable electrode contact. The integration of electronics with conductive yarns requires extra mechanical components to terminate the conductive yarn in a durable manner. Consequently, the manufacturing of devices by other means may be simpler and more cost-effective. The following section explores possible fabrication techniques for wearable health monitoring systems and smart textiles in general.

2.4 Fabrication techniques for textile-based wearable monitoring

This section examines fabrication techniques for wearable biopotential monitoring systems on textiles. The textile electronics fabrication techniques used in this thesis are stencil printing and screen printing. This section also discusses the suitability of inkjet and dispenser printing for textile electronics fabrication. Conductive yarns are examined as these are used to electrically connect components in all existing wearable monitoring systems. The final part of this section looks at other novel fabrication techniques, including photolithography.

2.4.1 Screen and stencil printing

Screen and stencil printing techniques involve depositing a paste on to a substrate by pushing it through a screen or stencil with a defined pattern. The difference between a screen and a stencil is that a screen is made of a fine mesh, which the paste must pass through when deposited, while the pattern gaps in a stencil have no mesh. A stencil and a screen for use in a DEK 248 semi-automatic screen printer are shown in Figure 2.4-1 and Figure 2.4-2 respectively.
Consequently, there are fewer constraints on the properties of the paste in stencil printing than screen printing. Functional pastes used in mesh screen printing must have functional particle sizes smaller than the gaps in the mesh, which usually limits the particles to less than 50 µm dependent on the screen mesh density and wire thickness. The bottom limit of viscosity for screen printing is around 1 Pa.s, whereas a higher viscosity is generally required for stencil printing so that the thicker structures that are produced hold their shape [37]. The top limit of the viscosity is also lower in screen printing than in stencil printing. Screen printing is described variously as having a top limit of 30 [38] or 50 [37] Pa.s whereas stencil printing can be used for any material that can physically be inserted into the stencil.

A stencil can print far thicker layers than a screen and screens are more appropriate where very thin layers are required. A layer from a screen print has a thickness equal to the thickness of the emulsion on that screen, usually no greater than 100 µm but ranging down to thicknesses of 5 µm. Stencils, on the other hand, rarely have a thickness less than 100 µm.

In most cases, a curing process is required to dry and solidify the material, changing it from a viscous paste to a solid layer. This is usually carried out with the exposure of the material to higher temperatures, although UV curing is also used.

2.4.2 Inkjet and dispenser printing

Inkjet printing uses pressure to jet droplets of ink out of a nozzle onto a substrate. The nozzle can move in the x and y directions relative to the substrate, often the substrate is moved and the nozzle is kept stationary [39], and is controlled by a computer system. In most cases a graphics package is used to make the design and the inkjet printer software translates this into movement instructions. The parameters that can be controlled, such as nozzle movement speed, droplet rate and number of cycles, are often varied to achieve different results. For example, Sawhney et al [40] printed a silver nitrate solution using 5 cm/sec, 500 drops/sec for 500 cycles.
Inkjet printer heads pump the ink either thermally or, more commonly, by using a piezoelectric membrane. This allows precise control of the flow rate. Inkjet printing flow rates are on the order of nL/min or pL/min. The variety of inks than can be used with this method is limited by ink viscosity and the nozzle size, which limits the size of particles in the ink. The permissible range of viscosities are different with different inkjet printers, the Fujifilm DMP-2831 in our laboratory has viscosity limits of 2-30 mPa.s according to the manufacturer datasheet [41]. Nozzles usually have a diameter of around 0.1-10 µm which limits the particle sizes accordingly. Inkjet printing is very useful for creating highly detailed patterns of a very low thickness.

Dispenser printing works in a similar fashion to inkjet printing; however the levels of ink flow can be several orders of magnitude larger. The nozzles are also larger, ranging from hundreds of µm up to cm in diameter, and can accordingly accept higher viscosities and particle sizes in the inks. The pumping mechanism for printed inks is pneumatic, so there are greater levels of ink flow but less precision in the exact volume deposited. Dispenser printing has flow rates generally measured in µL, several magnitudes above inkjet printing. Dispenser printing is therefore less capable than inkjet printing of creating low thicknesses and high resolutions, but useful when resolution can be sacrificed in favour of faster production of large areas or thick structures. Two implementations of dispenser assemblies used by Geng et al [42] are shown in Figure 2.4-3.

![Dispenser Assembly Diagram](image)

**Figure 2.4-3:** Two implementations of dispenser assemblies employed by Geng et al [42].

In conclusion, dispenser and inkjet printing are useful deposition techniques for fabricating high resolution and low thickness structures. Inkjet printing has better resolution than dispenser printing, but also more constraints on ink viscosity. These techniques are not appropriate on rougher substrates as the thickness of structures is low and consequently substrate imperfections are less tolerable.

Although dispenser and inkjet printers can be set up for relatively high flow rates, using multiple nozzles with wider diameters, this reduces resolution. With existing technology, a dispenser setup approaching the same speed of deposition as screen printing would necessarily have poorer resolution than screen printing.
2.4.3 Conductive yarns

Conductive yarns have been used in engineering research for at least 80 years [43] due to their textile compatibility and high durability. Conductive yarns are now used in most smart textile applications for connecting electrical components. Conductive yarns can be composed of exclusively conductive fibres, a mixture of conductive and textile fibres, or use a textile yarn coated with a conductive material [36]. Some coatings are applied by passing the yarn through a liquid or paste composed of the conductive material; however they can also be applied by wrapping a metal foil around the yarn. Adding insulation, so that conductive yarns can cross paths without shorting, will increase their thickness. Creating a yarn that is low resistance, thin, flexible, durable and electrically insulated is challenging. In most applications one or more of these properties must be sacrificed.

Conductive yarns are applied by sewing, stitching or embroidering them in to a textile. They have high durability but are difficult to connect to typical electronic components, especially if this connection also needs to be durable. Connections to components by conductive yarn is problematic because each yarn has to be stripped of its insulating coating, by a mechanical stripper or air blowing, and connected individually to hard electronics [44].

Conductive yarns have been used for skin contact electrodes by several researchers, such as Mestrovic et al [45] and Silva et al [46]. Some knitted yarn electrodes examined by Mestrovic et al are shown in Figure 2.4-4.

![Figure 2.4-4: Some knitted yarn electrodes examined by Mestrovic et al. (a) uses stainless steel yarns, (b) uses silver coated nylon, and (c) and (d) use silver/copper fibres blended with wool and polyester respectively [45].](image)

2.4.4 Other fabrication techniques

There are several other fabrication techniques that have been reported for fabricating textile electronics. These are described here.
The University of Ghent have developed and evaluated several processes that can be used to integrate electronics and stretchable connections on to textile. These involve either photolithography or laser cutting to create a conductive pattern from a thin sheet of copper or other conductor, which can have components attached and be encapsulated in a polymer. The textile integration involves printing a polymer in the required pattern and then placing the encapsulated electronics module on top, to bond with the textile printed polymer as it cures. This is reported with polydimethylsiloxane (PDMS), a type of silicone rubber, as the encapsulating polymer [47]. The resulting samples are classified by experimentation as waterproof but not machine washable. This fabrication method is complex but allows easy integration of electronics.

Sputtering has been examined as a method of creating conductive electrodes on textile. Jang et al [48] used Nylon laminated with polyurethane as a substrate for sputtering copper for ECG electrodes. They sputtered the polyurethane-coated side for 20 minutes, creating copper layers on top of the polyurethane lamination with thickness 2 µm. This layer had sheet resistance 0.316 Ω/square. The layer is flexible but this deposition process is time consuming in comparison to a screen printed flexible conductive layer, which can be deposited in a few seconds and then cured as part of a continuous process. Sputtering requires 20 minutes of deposition time for each sample. Further, the copper will oxidise and is not biocompatible, and it is noted in the conclusion that other materials must be explored if an ECG electrode is the aim.

Lamination is commonly used by research groups that use deposition processes to create textile based conductive paths and electrodes. Lamination is a way of significantly reducing the surface roughness of a textile with a minimal change in the flexibility of the textile. The laminate is heated so that it melts into the textile and compressed to ensure the polymer layer is flat when it cools. The result is a firmly bonded polymer layer on the textile with a flat surface. The textile retains its flexibility but the breathability is reduced. Many researchers have used lamination to allow them to deposit thinner layers of functional pastes or to improve print definition. Lamination can also be used on top of functional pastes as a protective layer. Kazani et al use an 80 µm thick polyurethane layer to encapsulate silver conductive pastes printed directly on textiles and wash them 20 times at 40 ⁰C, with no other items in the drum [49]. This improves their durability, and two printed textiles composed primarily of polyester show almost no change in resistance after 20 washes. However, because the silver is printed directly on to the textile 5-10 print passes must be used to fabricate a conductive track because the silver paste is absorbed by the textile. The thickness of the screen emulsion and the resulting silver layer is not disclosed.

2.4.5 Flexible electronics integration

This thesis focusses on the fabrication of passive components. Electronics integration is examined here as it is important for further development, allowing the production of active or capacitive electrodes. Screen printed tracks are easier to integrate with planar electronics than conductive yarns, as the components can be connected using traditional PCB manufacturing techniques. Here, several processes that have been introduced recently for the integration of
electronics into a flexible system are examined to provide guidelines for the integration of components into a screen printed system.

<table>
<thead>
<tr>
<th>Project</th>
<th>Research Group</th>
<th>Fabrication Technology</th>
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<td>Italian consortium</td>
<td>Conductive Yarn</td>
<td>2003</td>
<td>[33], [34]</td>
</tr>
<tr>
<td>Active electrodes</td>
<td>North Carolina State University</td>
<td>Screen Printing</td>
<td>2007</td>
<td>[27]</td>
</tr>
<tr>
<td>Flexible-Stretchable</td>
<td>Ghent/IMEC</td>
<td>Various</td>
<td>2011</td>
<td>[50]</td>
</tr>
</tbody>
</table>

The MagIC system (Italian consortium):

Although it has already been examined in section 2.3.3, the electronics in the MagIC system are discussed here to give an example of non-integrated electronics. In this system, the electronics are in a rigid case, which is attached to the conductive yarn system and detached during washing. The conductive yarn passive system can then be washed independently. There are advantages to rigid, detachable electronics, in that there is no need to fully waterproof the electronics and several garments can be used in conjunction with a single set of electronics. However, there are also disadvantages to this method. First, the rigid electronics, however small, could cause discomfort to the user when worn, especially during sleep. Second, this can only be used with passive electrodes, if active electrodes are used electronics at the electrode site are necessary and consequently must be integrated with the garment. Finally, it is possible for users to accidentally misuse the device, by washing with the electronics attached or by forgetting to reattach the device after washing. The detachable electronics module from the MagIC project is shown in Figure 2.4-5.

Figure 2.4-5: MagIC system with detachable electronics module above the right hip [4].
Printed active electrodes (North Carolina State University):

Kang et al [27] explore two fabrication processes to fabricate active electrodes on to Evolon, a non-woven textile. Both methods implement the circuit design for an active electrode shown previously on page 14.

The first method begins by printing a silver layer, as the skin contact electrode, on one side of the textile. The active electrode circuit is printed on the other side. The screen printed silver conductive paste used for these purposes is CMI 112-15. A hole is drilled in the textile and filled with conductive adhesive, CMI 119-05 and CMI 119-44 mixed at 100:2.3, to create a via that connects the electrode with the printed circuit. The components are attached to the printed circuit with the stencil printing and same conductive adhesive that was used for the via. This is then encapsulated in dielectric paste, CMI 112-30, using a dispenser printer. Carbon rubber electrodes are then attached to the top electrode using a conductive silicone bonding agent and sewn to the Evolon to ensure a secure attachment. The finished electrode is shown in Figure 2.4-6.

![Figure 2.4-6: Front (right) and back (left) of active electrode fabricated using the “direct attachment” method [27].](image)

The second method differs from the first because it uses a prefabricated active electrode circuit, or interposer, which is much smaller than the full screen printed electrode. This is still attached and encapsulated in the same way as individual components. However the power lines and signal out from the circuit, to connect to central electronics, are implemented with conductive yarns. This removes the need to encapsulate these connections and improves their durability. The finished electrode is shown in Figure 2.4-7.
Both types of electrode are washed 5 times in an unspecified wash cycle. The second design is shown to be more durable than the first as it is still functioning after the five washes, but it is acknowledged that it would be more problematic to mass produce due to the combination of fabrication techniques.

This work highlights the comparison between printed and conductive yarn textile electronics. The conductive yarns are more problematic when greater complexity is introduced, and the fabrication of the printed version is simpler. However, conductive yarns have greater durability than printed circuits on textiles.

Kang et al demonstrate that active, dry electrodes can have comparable performance to and greater durability than commercial Ag/AgCl electrodes. However, there are several limitations to the fabrication techniques described. The design requires deposition processes on both sides of the textile and a via through the textile. It is fabricated on a non-woven textile which is rarely used in garments worn next to the skin.

Flexible stretchable electronics (University of Ghent):

Vervust et al [51] describe a system in which the electronics are integrated into small, rigid, functional islands connected by flexible interconnections. They describe manufacturing techniques based on photolithography and laser cutting, which are both used to create meandering conductive patterns which can have components placed on top. These are then encapsulated on both sides with a polymer layer, with PDMS used in the photolithography process and an unspecified polymer used in the laser cutting process. These techniques are shown to create working devices with high water resistance, but are not tested by washing. The principle of “flexible-stretchable electronics” is shown in Figure 2.4-8.
Loher et al [52] describe a track layout also referred to by Vervust et al for stretchable systems. By printing meandering conductive tracks on to stretchable, non-woven substrates they were able to create stretchable conductive connections. In these structures the conductive material bends rather than stretching so that the whole structure can elongate in the required direction without a change in resistance or failure of the structure. This is used to connect rigid functional islands, and encapsulated in PDMS. The meander pattern is defined by the angle formed by interlocking arc segments, as shown in Figure 2.4-9.

Researchers have also reported electronic components specifically designed for flexible applications. Ultra-thin chip packaging (UTCP) is a way of fabricating integrated circuits so that they have low thickness and some flexibility. Sterken et al [50] report an ECG patch using UTCP chips with thickness 75 µm. The flexible patch has thickness 1.6 mm and a weight of 6.5 grams. This includes the power storage but not the electrodes. A flexible circuit comprising meandering conductive tracks and UTCP chips is shown in Figure 2.4-10.
2.4.6 Conclusions

It is obvious that the connection of an electronic component to a printed track is easier than connection to a conductive yarn. Electronic components are designed for connection to a flat layout and printed circuits on textile can have components attached in a similar manner to standard PCBs. The other technologies discussed, using photolithography or laser cutting and encapsulating in PDMS, is a viable solution for flexible electronics. It has been shown that this technology can be integrated with textiles by printing a layer of PDMS on to the textile, and bonding this with an existing system. The manufacturing complexity is, however, significantly higher than screen printing. The UTCP chip packaging, and other, similar developments, will improve the capability and durability of flexible electronic systems. This technology appears to be compatible with both screen printed and photolithographic flexible electronics production techniques but would be more difficult to integrate with conductive yarns.

Out of the thick and thin film deposition processes discussed, it is clear that the most appropriate process for the types of structures required for biopotential monitoring networks on textiles is screen printing. It has the appropriate volume of deposition, layer thickness and resolution for the surface roughness of the textile and the structures required. This is supported by the evidence that screen printing is currently the favoured technique for printing directly on to textiles in the fashion industry, although inkjet printing is used to produce finer detail on top of this layer.

<table>
<thead>
<tr>
<th></th>
<th>Volume deposited</th>
<th>Resolution</th>
<th>Ink/Paste Viscosity</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stencil</td>
<td>High (mL/min)</td>
<td>Low (mm)</td>
<td>High (Paste)</td>
</tr>
<tr>
<td>Screen</td>
<td>/</td>
<td>/</td>
<td>/</td>
</tr>
<tr>
<td>Dispenser</td>
<td>/</td>
<td>/</td>
<td>/</td>
</tr>
<tr>
<td>Inkjet</td>
<td>Low (pL/min)</td>
<td>High (nm)</td>
<td>Low (Ink)</td>
</tr>
</tbody>
</table>

Inkjet and dispenser printing would be useful techniques to integrate eventually, as they would allow finer resolution and lower layer thickness for functional pastes, reducing cost. It would also allow small adjustments to be made to a circuit layer without the need to design and manufacture a new screen.
Conductive yarns remain the most common method of implementing textile-based electronics. Although this fabrication technique is relatively simple, it is very difficult to make connections to electronics so this technique is limited when the aim is to fabricate smart textiles with more complex functionality. Deposition methods are more appropriate for anything involving integrated components, such as passive and active circuits, or sensor structures, such as cantilevers.

Kang’s direct-attach method of including circuit elements in a printed textile is the most suitable method to integrate electronics components with screen printing. Given that there were problems with the durability of these electrodes, the design must be changed to fabricate an active electrode with improved durability while keeping the manufacturing process homogeneous.

2.5 Printed textile electronics

The final part of the literature review focuses on printed electronics on textile. The first section examines methods used to print conductive tracks and the second looks at printing to obtain skin contact electrodes for biopotential monitoring.

2.5.1 Printed flexible conductive tracks and durability

This section discusses research into printed conductive tracks on flexible substrates. There are multiple aspects to this research area, such as the behaviour of inks and pastes on porous substrates, methods of creating a smooth surface above the textile for higher resolution printing and methods of increasing durability and flexibility. The papers examined in this section are given in Table 2-3.

<table>
<thead>
<tr>
<th>Paper</th>
<th>Research Group</th>
<th>Printing Type</th>
<th>Substrate</th>
<th>Year</th>
<th>References</th>
</tr>
</thead>
<tbody>
<tr>
<td>Using conductive inks and non-woven textiles for wearable computing</td>
<td>North Carolina State University</td>
<td>Inkjet</td>
<td>Non-wovens (Tyvek, Evolon, RPM)</td>
<td>2005</td>
<td>[31]</td>
</tr>
<tr>
<td>Electrical characterization of conductive ink layers on textile fabrics: Model and experimental results</td>
<td>Barcelona Institutes</td>
<td>Screen</td>
<td>Polyurethane-coated polyester</td>
<td>2007</td>
<td>[54]</td>
</tr>
<tr>
<td>Electrical conductive textiles obtained by screen printing</td>
<td>Ghent University</td>
<td>Screen</td>
<td>Woven textiles</td>
<td>2012</td>
<td>[49]</td>
</tr>
</tbody>
</table>

Using conductive inks and non-woven textiles for wearable computing:

Kang et al describe research carried out in North Carolina State University on inkjet printing on to non-woven textiles. Both the substrate types and fabrication processes are different in this paper to those used in this thesis, however a detailed examination of substrate porosity and
ink penetration is provided. Many of the observations are relevant to experimental work in later chapters.

Two different conductive pastes were each inkjet printed on to three substrates. The pastes used were CMI 112-15 made by Creative Materials and CSS-010A made by Precisia. The substrates used were Evolon, Tyvek and Resolution Print Media (RPM) and they were cured at 140 °C, 110 °C and 160 °C respectively. The pastes were printed on to the substrate as co-planar waveguides as shown in Figure 2.5-1.

The waveguides were printed with transmission line width (a) of 0.6, 1.0 and 1.3 mm. Transmission lines 2 cm and 6 cm in length are examined and the resistance of these tracks does not exceed 7 Ω. Sheet resistances are not provided. Apart from measurement of the DC resistance, several other tests are performed. Their observations are as follows:

- DC resistance values were measured. CMI 112-15 had lower resistance values than CSS-010A as it had a larger proportion of silver. Printed conductive layers on Tyvek and Evolon had similar values of resistance while those on Resolution Print Media had almost twice the resistance. This is explained below.
- SEM imaging was used to examine the textile and ink height above the textile. Printed tracks on Resolution Print Media (RPM) had a height of around 20 µm above the textile while those on the other nonwovens had an height of around 40 µm above the textile. This is due to the porous structure of RPM, the ink penetrates into the material and so less remains on the surface. This correlates with higher resistance values for the RPM.
- Viscosity measurements were taken with a Brookfield Cone/Plate Viscometer, showing that CMI 112-15 has higher viscosity than CSS-010A. At a shear rate of 50 s⁻¹ CMI 112-15 has a viscosity of 21 Pa.s while CSS-010A has a viscosity of 9 Pa.s.
- Ink penetration is measured using ultrasound, with a greater intensity of reflected sound showing a greater penetration. The results show that there is a greater penetration of ink in RPM. Tyvek and Evolon have comparable values despite Tyvek having lower porosity.
- A high speed camera is used to examine the behaviour of individual droplets as they are jetted on to the textile surface. CMI 112-15 remains on the surface and penetrates less due to its greater viscosity.

For a conductive track it is desirable that the ink remains on the surface to ensure high conductivity. This paper shows the importance of factors such as ink viscosity and substrate
structure when controlling the penetration of the ink. High ink penetration reduces the conductivity of the resulting layer.

Electrical characterization of conductive ink layers on textile fabrics: Model and experimental results:

Rius et al investigated modelling the resistance of screen printed conductive inks on textile substrates. A theoretical model for the resistance between two points in a conductive plane was created. This model was tested with printed planes on a textile to gauge the predictability of their electrical properties. Acheson Electrodog 965SS carbon ink was screen printed on to a knitted textile substrate with a laminated polyurethane coating. The printed areas were rectangles with dimensions 290 x 195 mm. The model was an accurate fit when the characteristics of the printed rectangle were uniform. When the model did not fit it could be explained by a poor quality print. Rius et al also note that the model correlates less well with experimental results as the distance between the two points increases, as all the imperfections in the plane accumulate. From this it can be seen that it is important to minimise imperfections on the substrate surface for any conductive track where the resistances of multiple batches must be repeatable.

Electrical characterization of screen printed circuits on the fabric:

Yoo et al evaluate screen printing on to textiles. Silver polymer paste Changsung Corporation CSP-3163 is screen printed on to polyester with a screen emulsion thickness of 66 µm, resulting in a cured layer around 40-50 µm thick. An SEM micrograph of the cross section is shown in Figure 2.5-2.

![Figure 2.5-2: An SEM micrograph of screen printed silver ink on polyester [56].](image)

Conductive tracks are printed on to polyester with widths of 0.2, 0.5 and 1 mm and length 20 mm. Some printed tracks are coated in an unspecified passivation layer, and all samples are washed. The wash process used is unspecified. It was found that the coated conductive tracks had low resistance after 50 washes, while the uncoated tracks had significant resistance increases. The normalised resistances over 50 washes for four tested conductive tracks, two with and two without passivation, are shown in Figure 2.5-3.
This paper also examines the use of a hard epoxy coating to protect microchips which are wire bonded to screen printed bonding pads. In tensile tests the textile ripped before the microchip was damaged, demonstrating the feasibility of this technique for encapsulating textile electronics. This work shows again that a coating on a printed conductive path significantly increases durability. As in the previous paper, a very thick silver layer was used, increasing the cost of the device. This work also demonstrates that bonding a microchip to a printed textile is significantly simpler than bonding conductive yarn.

**Electric conductive textiles obtained by screen printing:**

Kazani et al provide an approach to find which textiles give greater durability when they are used as substrates for screen printed conductive layers. In this paper two silver pastes, Electrotag PF 410 and Dupont 5025 Silver Conductor, are screen printed in a square measuring 60 x 60 mm on to a series of textiles. The thickness of the silver layer is not given but based on the Ω/square resistances achieved, and the fact that there were 5 to 10 print passes in each case, the layer thickness is likely to be around 50 µm. These samples were washed 20 times. It was found that resistance increased greatly due to cracking and in half the cases conductivity was lost completely. The textile was more important than the paste in dictating the durability. They note that high absorption of the textile reduces the thickness of the printed silver and improves the durability. They also note that fabrics containing polyester performed well as they suffered less plastic deformation.

They then cover newly printed samples of eight of the most durable textiles in the previous test in a polyurethane sheet, Epurex LPT, of thickness 80 µm. This is melted on to the textile surface on top of the printed square. This method is successful in maintaining a low resistance in the printed silver with resistivity below 0.2 Ω/sq after 20 washes. The best results were obtained with fabrics PES 4 (polyester) and PES/CV (polyester/cotton viscose). The sheet resistance with a Dupont 5025 layer increased from 0.008 to 0.010 Ω/sq on PES 4 and 0.027 to 0.028 Ω/sq on PES/CV. The results with PF 410 were similar.

This paper shows that protecting printed silver polymer tracks with polyurethane is effective in increasing the wash durability in smart textiles. Since the polyurethane layer is laminated it will
adhere to and hold the silver layer and textile in place, but the polyurethane layer reduces the textile flexibility. Screen printed polyurethane is preferable to laminated polyurethane because screen printing allows the pattern to be printed only where required, maintaining the flexible and comfortable feel of the textile, whereas laminating generally covers the whole textile.

Conclusions:

These four papers provide several pieces of information which will aid the production of screen printed conductive tracks on woven textiles. The most durable printed conductive textiles are obtained with an encapsulation layer. This suggests it is the best method to make durable functional textiles. As well as protecting the conductive layer from moisture and abrasion damage, the addition of an encapsulation can reduce the strain on the conductive layer during deformation by moving it closer to the middle of the structure where there is less dimensional change during flexure.

Conductive tracks of length 1 m and width 2 mm would require 500 squares of conductive material, which is considered in this thesis to be the maximum number of squares for a conductive track used in a biopotential monitoring garment. The resistivity of the silver pastes used in the described literature (<0.1 Ω/sq before washing) would provide sufficiently low resistance tracks for biopotential measurements based on the previously defined limit of 100 Ω. The textiles used in the described papers, whether woven or non-woven, generally showed some absorption of the silver ink, requiring printers to use a large amount of silver ink. As this ink is expensive, a polymer interface printed on the textile which provides a smooth surface for printing conductive layers would be useful to reduce the required amount of silver paste used.

Various methods of washing are used by institutions to determine the durability of a particular technology. These range from bain-marie washing, where a single sample is submerged in water in a slowly turning container, to more harsh wash cycles, sometimes referred to as industrial washing, with temperatures of 80 or 90 °C and spin speeds of 1000 rpm or more. Printed conductive textiles tested in the literature are generally washed with an empty washing machine. The author’s experience has shown that the presence of other textiles in the machine affects the number of washes before failure because it increases the stress on the printed conductive textile.

The project described in this thesis aims to create printed conductive textiles with a thin (>10 µm) silver layer deposited in a single print, which is durable to 10 standard washes once cured. A standard wash is defined here as a wet wash in a washing machine with other garments at 40 °C, with a maximum spin speed of 1000 rpm for 60 minutes.

2.5.2 Printed skin contact electrodes for biopotential measurements

This section examines various electrode designs that incorporate screen printing as the fabrication technique, or as one of several fabrication techniques. This review contains five examples of research selected because of the similarity of the reported technology to the technology that is developed in this thesis. The five research projects examined here are given in Table 2-4.
Table 2-4: Printed textile electrode research reviewed in this section.

<table>
<thead>
<tr>
<th>Paper</th>
<th>Research Group</th>
<th>Active/Passive</th>
<th>Substrate</th>
<th>Year</th>
<th>References</th>
</tr>
</thead>
<tbody>
<tr>
<td>NIBEC ECG mapping harness</td>
<td>NIBEC</td>
<td>Passive</td>
<td>Polyester film</td>
<td>1992</td>
<td>[57]</td>
</tr>
<tr>
<td>Sensory baby vest for the monitoring of infants</td>
<td>Institute of Textile Technology and Process Engineering</td>
<td>Passive</td>
<td>Textile (Woven)</td>
<td>2006</td>
<td>[26]</td>
</tr>
<tr>
<td>Fabric-based active electrode design and fabrication for health monitoring clothing</td>
<td>North Carolina State University</td>
<td>Active</td>
<td>Textile (Non-woven)</td>
<td>2008</td>
<td>[27]</td>
</tr>
<tr>
<td>Electrical characterization of screen-printed circuits on the fabric</td>
<td>KAIST</td>
<td>Passive</td>
<td>Textile (Unspecified)</td>
<td>2009</td>
<td>[55]</td>
</tr>
<tr>
<td>Smart jacket for neo-natal monitoring with wearable sensors</td>
<td>Eindhoven University of Technology</td>
<td>Passive</td>
<td>Textile (Woven)</td>
<td>2009</td>
<td>[58]</td>
</tr>
</tbody>
</table>

**NIBEC ECG mapping harness:**

The earliest example of a screen printed system for monitoring human biopotentials is the electrode technology fabricated in the Northern Ireland Bio-Engineering Centre (NIBEC) [57]. Although this is not a textile system, its description provides useful information on the fabrication of a flexible multiple electrode systems for ECG. This system was designed to take detailed heart examinations without needing to be in a hospital, allowing the correct treatment for a given heart condition to be provided more quickly.

Their electrodes are comprised of a screen printed Ag/AgCl paste on a 125 µm thick polyester substrate. A hydrogel was applied by hand to the electrode pads. McLaughlin et al note that this results in a higher electrode resistance than normal wet Ag/AgCl electrodes due to a lack of hydration of the skin, resulting in further impedance issues. They suggest, however, that the advantages of hydrogel electrodes over wet electrodes outweigh the impedance considerations, noting the simple sensor design and manufacture, flexibility, relatively thin structure, good adhesion, and lack of shorting between electrodes. The total thickness of the substrate, printed pads and hydrogel was 700 µm. The system can be recycled by disposing of the hydrogel pads and applying new pads to each electrode.

In order to fabricate a high density ECG monitor, a printed system on polyester is cut into what are described as ‘fingers’ so that different body sizes and shapes can be easily accommodated. This allows a variation in spacing between electrodes, reducing the movement of the electrodes themselves and thereby reducing the severity of motion artefacts on the resultant signal. The NIBEC system shown in Figure 2.5-4 has four limb electrodes and 64 chest electrodes.
This paper demonstrates that printing is a viable technology for low-cost production of flexible biopotential monitoring networks. It is shown that printed silver tracks can be used to connect skin-contact electrodes to electronics, and that this can be used by physicians. It is noted that the hydrogel electrodes had to be replaced for each use. To make a more durable device neither wet electrodes nor hydrogel electrodes can be used.

**Sensory baby vest for the monitoring of infants:**

Linti et al [26] fabricated a system for infant monitoring, in which silicone rubber loaded with silver particles is used as the electrode material. The type of printing used is not stated in this paper. Given the thickness of the electrode which appears to be around 250 µm in provided images, stencil printing was probably the manufacturing technique used. They also use carbon-loaded silicone rubber for strain gauges in the monitoring of breathing. Carbon loaded silicone rubber has a greater resistance than silver loaded silicone rubber so there is less power consumption when it is used as a resistive strain gauge. The sensor components are connected to central electronics by Teflon-coated AWG 36 wires.

This paper demonstrates that conductor-loaded silicone rubbers can be used as an electrode material, being more flexible and more resistant to moisture than most commercial silver pastes. The low cost of formulating a conductor filled silicone rubber makes it an attractive alternative to commercial conductive polymer pastes, especially when the layer thickness is increased above 100 µm. An SEM micrograph showing the cross-section of the silver-loaded rubber is given in Figure 2.5-5.
Fabric-based active electrode design and fabrication for health monitoring clothing:

The electrode designed by Kang et al was described earlier as an example of electronics integration. Here, the performance and materials of the electrodes fabricated with the printed (or “direct attachment”) method are discussed. CMI112-15 silver polymer paste is printed on to Evolon and cured at 100°C. As with Linti’s work, conductor loaded rubber is used as the skin contact part, but this is not directly deposited on to the textile. Instead, prefabricated carbon loaded rubber electrodes are bonded to the silver-printed surface using conductive silicone loaded epoxy. Clearly this bonding strength is not considered sufficient because the carbon loaded rubber is then sewn to the printed textile. A via through the textile connects to the active electrode circuit. The improved performance with the active electrodes compared to passive electrodes with the same structure is clear during jogging, where the susceptibility to motion artefact is greatly reduced, as shown in Figure 2.5-6.
Electrical characterization of screen-printed circuits on the fabric:

Yoo et al [56] print electrodes directly on to textiles for the monitoring of ECG. The pastes used are Changsung corporation CSP-3163 and Fusikara Kasei corporation Dolite D-500, both silver polymer pastes, although no comparison between the pastes is provided. They print silver paste in a thickness of 10 to 30 µm in a circle with diameters from 10 to 30 mm. The method of connection of the textile electrodes to hard electronics is not described. The system is printed on to an inelastic textile which is subsequently attached to an elastic textile for good conformance to the body.

In a comparison between the measured impedances of printed textile electrodes of 10, 20 and 30 mm diameters a significant difference between skin-electrode impedance levels was observed. The silver paste on textile is shown in Figure 2.5-7 and the impedance comparison is shown in Figure 2.5-8.

Yoo et al conclude that their system is suitable for long term monitoring because it uses dry electrodes. Their method of printing on to an inelastic textile and then attaching this to an elastic textile provides mechanical support to the printed structure while allowing their electrodes to conform to the body.

Smart jacket for neo-natal monitoring with wearable sensors:

Bouwstra et al [58] state that critically ill new-borns are very sensitive to external disturbance; constantly attaching and removing various electrodes and other monitoring devices can be detrimental to the infant’s health. They proposed a wearable system to monitor infant ECG.

Two types of electrode are fabricated. One is a textile printed with a gold paste developed at the Netherlands Organisation for Applied Scientific Research, and the other is an electrode woven with silver plated yarn manufactured by Bouwstra et al. These electrodes are integrated with several layers of cotton for comfort. The electrodes, connections and cotton were sewn together using Shieldex silver 117/17 (82 Ω/m) and Shieldex silver 235/34 2-Ply (100 Ω/m) conductive yarns. The two different yarns appear to be used for different mechanical properties.
The system developed uses six electrodes, but does not use all six simultaneously. It measures which electrodes have the best contact by observing signal strength. The electrodes with the best contact will change as the position of the infant changes, because the pressure on the electrode affects the signal quality. The system was able to consistently maintain a stable electrode contact and an artefact-free ECG wave, and Bouwstra *et al* also assert that it is reasonably comfortable.

They conclude that different electrode designs work well with different materials. Gold printed electrodes with a diameter of around 15 mm and silver woven electrodes with a diameter of around 40 mm both gave a good ECG signal with low noise. This is stated as being due to the different electrical resistivity of gold and silver, however there is also the factor of the rougher surface of the textile electrode, reducing the skin contact area. The printed and woven electrode types both have associated disadvantages; the gold printed electrodes deteriorate with washing and the silver woven electrodes require pressure to be applied to give a stable signal. The electrodes are shown in Figure 2.5-9.

![Textile electrodes fabricated by Bouwstra *et al*.](image)

This paper demonstrates that comparable performance with woven electrodes can be achieved with smaller printed electrodes because the conductivity of the electrode is higher and the surface is smoother allowing a larger greater skin contact area for the size of the electrodes. The permissible electrode density in a printed system is therefore greater than in a woven system for a given signal to noise ratio.

**Conclusions:**

In this thesis a homogenous production process is required, to allow electrodes and conductive paths to be integrated into a textile using only printing fabrication processes. This would provide a method for easy fabrication of complex flexible networks of electrodes for specific biopotential sensing functions, such as EEG networks. Although various printed electrode designs have been used successfully, there is not yet an example of a fabrication process that allows both conductive paths and electrodes to be integrated into textile using a deposition-based technique such as printing.
2.6 Conclusions

The aim of this project is to create a process for fabricating biopotential monitoring hardware on textiles. This hardware will include electrodes and conductive paths. Dry electrodes must be used for this to be appropriate for durable long term monitoring. Some researchers have described impedance issues with passive dry rubber electrodes so active electrodes will also be examined.

Screen printing will be used to fabricate electrodes and conductive paths. Printing is a relatively unexplored fabrication technique for biopotential monitoring. However, research in which printing is used for this purpose shows that there are several advantages over conductive yarn, such as compatibility with planar electronics, appropriateness for continuous fabrication and smoother electrode surfaces which provide larger skin contact areas. Of the printing processes, screen printing is the most appropriate for the fabrication of structures on textiles however inkjet and stencil printing would be useful for finer detail and thicker layers respectively.

The fabricated systems must be durable. The examined literature shows that encapsulating printed conductive pastes with a polymer increases their durability. Encapsulation prevents abrasion to the silver and also moves the silver closer to the middle of the structure, where there is less dimensional change. A polymer interface layer printed before the conductive paste would also be beneficial for two reasons; the conductive paste would be protected from moisture and abrasion from both sides, and the interface layer would reduce the absorption of the silver paste. The aim in this thesis is to produce conductive tracks that are durable for 10 standard domestic machine washes with a silver layer under 10 µm thick.
3 - Design Methodology

3.1 Introduction

Given the requirements of a wearable health monitoring system and the constraints of screen and stencil printing, several design rules must be defined. This chapter firstly examines the way in which a textile-based monitoring system must be designed using screen printing technology. This is followed by a section discussing the selection of appropriate textiles for this work. After these the general design issues surrounding printed conductors on textiles are discussed, drawing information from the literature review and justifying the approaches used in later chapters.

This chapter gives detailed designs for textile printed conductive paths, skin contact electrodes and via connections. The descriptions of these fundamental designs provide a framework for the selection of materials in the following chapter.

3.2 Requirements for a wearable biopotential monitoring system

This section discusses the design requirements. The basic requirements for a functioning wearable biopotential monitoring system are described first, then the constraints imposed by the screen printing fabrication method are examined.

3.2.1 Functional requirements of a wearable biopotential monitoring system

The fundamental requirement for a system measuring ECG from several angles, such as the Frank configuration vectorcardiogram (VCG), is to have sensors dispersed over the body and connected to a central location. The positions of these sensors on the skin surface should not change significantly during movement as this will cause motion artefacts on the acquired ECG signal. This applies equally to other biopotential monitoring situations but is especially relevant on the torso as there is significant dimensional change due to the contraction of skeletal muscles.

Preliminary work with flat, dry electrodes mounted on textiles indicated that the electrodes should protrude from the textile surface by around 1-3 mm to ensure robust contact. Maintaining stable contact between the electrodes and the skin is an important issue as the electrodes have no self-adhesive properties. Adhesive electrodes are avoided in this thesis as they have low durability; hair and dirt gradually gathers on the adhesive, reducing conductivity by reducing skin contact area, and preventing the adhesive from functioning.

These electrodes must conform to the skin to maximise the contact area between the skin and the electrode, thereby minimising the skin-electrode resistance. The resistance of the electrode should be less than 1 kΩ, so that it does not contribute significantly to the overall resistance. It is important that the variation between the resistances of different electrodes is
minimal so that any noise couples equally into both amplifier inputs and is rejected as common mode.

Some stretchability in the wearable network would allow garments to be stretched slightly when worn, increasing contact stability of electrodes during body movement. Stretchability will also allow a garment to be stretched over different body sizes without affecting the placement of electrodes relative to the heart. This will reduce the number of different garment sizes required to cover a given distribution of body sizes and shapes.

3.2.2 Design requirements for compatibility with screen printing on textiles

There are several constraints that are imposed on the design of the device by the screen printing technique. The first is surface roughness. Excessive surface roughness will make a textile problematic to print, and it is difficult to screen print across seams. This constrains the layout of a device, and the textiles that can be used. It is necessary for each printed connection between electrodes and central electronics to be fabricated over a single piece of woven textile, as seams between textiles disrupt a printed track. Most garments that cover the torso are composed of two pieces of textile, a front and back, that are sewn together to form a vest with a seam at either side. This will not be possible in a Frank configuration garment as the printed conductive tracks will need to reach all around the torso, including an electrode on the back.

An electrode thickness of from 1 to 5 mm is required to provide a stable skin contact. This thickness is time-consuming to obtain by screen printing, where deposits are normally limited to a maximum of 100 µm. Consequently it was decided that this layer should be stencil printed rather than screen printed, so that this thickness can be obtained in a single printed layer.

To increase the range of textile surface roughness that can be printed on, a polymer interface layer developed in the University of Southampton [59] is used in this work. This creates a thin, flexible film on the surface of the textile with a smooth surface. Subsequent layers can be more easily printed on to this interface layer.

Excessive extension will break screen printed connections on a textile. However, some stretchability is necessary in order to allow the garment to conform to the skin without being constrictive to the user. Because of these contradictory stretchability requirements, work in this thesis is carried out on an inelastic printed textile, which is subsequently sewn on to an elastic garment. The garment is then able to stretch and conform to the body without straining the functional components. This is the same methodology employed by Yoo et al [56].

3.3 Textile selection and characterisation

The selection of elastic and inelastic textiles is required. The most important factors in this selection are the textiles’ mechanical properties but there are several other requirements. A summary of the proposed requirements for the two textiles is given below. This summary serves as a starting point for the selection of textiles.

For the inelastic, printed textile:
- The textile must be breathable and washable.
- The textile must have a maximum extension of less than 20% during wearing and washing to prevent damage to printed tracks.
- The textile must have a thickness less than 0.5 mm.
- The textile must be able to withstand temperatures of 130 °C to allow the heat curing of conductive pastes.

For the elastic textile used as a backing for the printed textile:
- The textile must be breathable and washable.
- The textile must have an elastic extension range of at least 80% to allow it to conform to the body.
- The textile must have a thickness of less than 1 mm.

The thickness requirements are not essential for the operation of the device, but keeping the total textile thickness less than 1.5 mm will minimise the thickness potentially making the device more comfortable to wear. With the printed structures fabricated in this thesis, usually below 500 µm, the total device thickness would be less than 2 mm, meaning it can be worn comfortably and fairly discreetly under normal clothing.

Two inelastic textiles from Klopman International [60], known as Escalade and Lagonda, one elastic textile from Elasta [61] and one elastic textile from a Fabricland [62] were all examined as candidate textiles for use in this work. The inelastic textiles were supplied on a roll of width 1.5 m.

The elastic textiles are in the form of ‘bands’, with many widths available. Widths of 50 mm (Elastic) and 60 mm (Elasta) are characterised here as these were appropriate widths for the sensing garments developed in this thesis. SEM micrographs, showing the weave of Escalade, Lagonda and Elasta are shown in Figure 3.3-1.

![Figure 3.3-1](image)

**Figure 3.3-1: SEM micrographs of the weave structure of textiles used in this thesis. The level of magnification is the same in each micrograph presented. Left: Lagonda, Middle: Escalade and Right: Elasta.**

Tensile measurements were taken from each textile by extending each sample until failure. A Tinius Olsen H25KS was used to tensile test samples of textiles. This is referred to in this thesis as the standard tensile test apparatus. The three characteristics extracted from these tests are:

- **Tensile strength**: The stress required to break the sample.
- **Young’s modulus**: The amount of stress required to deform the material.
- **Elongation at break**: The amount of extension when the sample is broken.
Unless stated otherwise, all tensile tests in this thesis extend samples until they break. More information on the extraction of these parameters and the tensile testing apparatus can be found in appendix B.

The tensile test samples had a gauge length of 70 mm for the inelastic textiles and 80 mm for the elastic textiles. The width of the samples was 40 mm for the inelastic textiles and 50 and 60 mm for the elastic textiles.

Figure 3.3-2 shows a tensile test on samples of Escalade and Lagonda in the warp and weft directions. Tensile tests on the textile in non-orthogonal directions (i.e. neither warp nor weft) resulted in an unravelling of the weave due to the comparatively small size of the sample. These results are omitted from the graph.

![Figure 3.3-2: Results of a tensile test of Escalade and Lagonda in the warp and weft directions.](image)

These results show that Escalade has a greater tensile strength and higher Young’s modulus than Lagonda. Using Escalade would make planar extension of the printed textile less severe, minimising the deformation of the printed track during a tensile force. There is also a clear difference between the tensile properties in warp and weft directions for both textiles. This should be considered when selecting the orientation in which the textile should be printed.

Figure 3.3-3 shows a tensile test with samples of elastic textiles Elasta and Elastic. These are tested in the direction of greater stretchability.
Figure 3.3-3: Results of a tensile test on Elasta and Elastic textiles.

Figure 3.3-3 shows that Elasta and Elastic textiles are very similar in terms of tensile properties, with Elasta having a slightly greater extension before break. Elasta also has a slightly lower Young’s modulus than Elastic. The linear-elastic portion of the curve lasts to around 120% extension for both textiles.

The thickness of samples and other investigated parameters for the textiles are summarised in Table 3-1.

Table 3-1: Measured parameters for four textiles used in this work.

<table>
<thead>
<tr>
<th>Textile</th>
<th>Escalade</th>
<th>Lagonda</th>
<th>Elasta</th>
<th>Elastic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Supplier</td>
<td>Klopman</td>
<td>Klopman</td>
<td>Elasta</td>
<td>FabricLand</td>
</tr>
<tr>
<td>Thickness (µm)</td>
<td>410</td>
<td>290</td>
<td>890</td>
<td>1035</td>
</tr>
<tr>
<td>Resistance to Temperature</td>
<td>Fine up to 130 °C</td>
<td>Fine up to 130 °C</td>
<td>Warping around 90 °C</td>
<td>Warping around 90 °C</td>
</tr>
<tr>
<td>Extension at break (%)</td>
<td>Warp: 48 Weft: 10.5</td>
<td>Warp: 13.5 Weft: 49</td>
<td>225</td>
<td>205</td>
</tr>
<tr>
<td>Young’s Modulus (N/cm)*</td>
<td>3.14</td>
<td>13.74</td>
<td>1.93</td>
<td>10.27</td>
</tr>
</tbody>
</table>

*For inelastic textiles the Young’s modulus at break is given. For the elastic textiles, the Young’s modulus is taken from the elastic (straight line) portion of the tensile characteristic at 100% extension.

Any of these textiles could be used based on the defined textile requirements. Lagonda and Escalade both have advantages. Lagonda is a thinner textile and has a finer weave, which provides a lower surface roughness and makes printing easier. However, Escalade has greater tensile strength and a higher Youngs modulus and may protect printed tracks from damage during planar tension more effectively. Elasta (Elasta) is preferable to the elastic (FabricLand) because it has a lower thickness and lower Young’s modulus. The low Young’s modulus will reduce the stress for a given extension and therefore reduce how constricting the device is to wear. Elasta will be used for the garments in this thesis.
3.4 Screen printed conductive tracks

This section discusses the design of printed conductive tracks on textile. First, the concept of an interface layer is introduced. The design of the printed tracks to maximise durability is then discussed. The layout of a network composed of these printed tracks is examined, and the design for printed track terminations is provided.

3.4.1 Interface and encapsulation layers

In many existing screen printed textiles, the conductive or active material is printed directly on to the textile. This technique tends to give poor durability, and it also greatly limits the allowable roughness of the material to be printed. Durability can be improved with a greater number of layers of conductive material, however this increases the amount of conductive paste which must be used. As low resistance conductive pastes contain precious metals, usually silver, the financial cost of the production technique increases significantly with extra layers of conductive paste.

In this work, the textile is printed with a flexible polymer interface. This creates a smooth surface on the textile so that a single deposit of conductive paste can be used to establish a conductive layer. The number of layers of polymer required to create a smooth surface is dependent on the polymer interface material and on the surface roughness of the textile. A diagram of an interface layer on textile providing a smooth surface for a thin conductive layer is shown in Figure 3.4-1.

![Diagram](figure3.4-1.png)

Figure 3.4-1: Cross-sectional diagram of an interface layer creating a smooth surface above a rough textile surface for the printing of conductive pastes.

As demonstrated in the literature review, an encapsulation layer of a dielectric polymer material can also be printed on top of the conductive layer to protect it from damage. This will prevent the ingress of water, protect from abrasion damage, and minimise the mechanical strain on the conductive layer by moving it to a plane in the structure where there is less deformation during flexure.
Figure 3.4-2: Cross-sectional diagram of a three layer conductive track structure on textile.

Because all layers are screen printed it is possible to deposit an interface layer only where necessary. The rest of the textile can then retain its flexibility and breathability. This fundamental design is used for all printed conductive tracks fabricated as part of this research.

3.4.2 Durability of conductive tracks

The literature review indicated that the durability of printed conductive tracks is one of the main weaknesses of printed systems compared with conductive yarn systems. There are four main approaches to increasing the durability of the conductive tracks:

- **Controlling the conditions to which the device is subjected.** These can be controlled with defined washing and storage procedures but it is expected that the textile will be subjected to significant mechanical stress and moisture during its usable lifetime. It is preferable to fabricate a device that does not require special washing conditions for two reasons. Firstly, users will inevitably wash the garment accidentally which will increase the cost of the deployment of this technology. Second, special washing requirements reduce the ease of use, which would be one of the main advantages over existing systems.
- **Increasing the tensile strength of the device.** The device can be made more durable by increasing the stiffness to the point where it is unlikely that there will ever be sufficient stress to extend printed structures. The effect of planar stresses is reduced by choosing an inelastic textile as the substrate, so that the extension even under high stress is low. The use of a polymer interface layer also increases the stiffness of the conductive track compared with an unprotected conductive layer. Increasing stiffness to the point where the printed conductive track cannot bend, however, will reduce the flexibility of the textile and thereby reduce the wearability of a device.
- **Increasing the flexibility of the device.** The flexibility of the device is crucial if the device is to be wearable. Bending of a printed textile normal to the planar structure of the device will deform the printed tracks. Elastic properties will allow the printed structures to extend without tearing and then return to their original shape when stress is reduced.
- **Device structure to minimise stress on functional materials.** The design of printed structures to minimise stress on functional materials will improve durability. Both the layer structure and the planar layout should be designed to minimise stress on conductive layers. This method is used in flexible displays, where the functional
photovoltaic materials are positioned carefully between two thicker sections of polymer [63]. The position in a structure where there is no deformation during bending is known as the neutral axis.

Methods for controlling the stress during wearing, washing and storage will not be considered in this thesis. Achieving durability by creating highly flexible structures is preferable to achieving it by creating very strong structures if a system is to be wearable. Some researchers have suggested that the optimal approach uses both techniques, with high stiffness, encapsulated islands for electronics, sensors and actuators, and flexible, stretchable conductive paths between them [64]. A further discussion of the tensile properties of materials and their effect on durability is given in chapters 4 and 6.

The design of the printed structures can also affect their durability. For example, sharp internal corners in a flexible system will cause a concentration of stress at the corner. Here, two ways of improving the durability of thick film structures by design changes are discussed. These are the horse-shoe layout, which allows the fabrication of stretchable conductive tracks with relatively stiff functional materials, and the placement of conductive materials in the neutral axis, which minimises the deformation of these conductive materials during flexure.

**Horse-shoe layout:**

Device layouts to minimise stress can be found in the literature review. Work by Loher et al [52] describes a horse-shoe layout for stretchable conductive tracks. As the substrate is stretched, its width decreases as its length increases. Consequently the distance covered by the copper foil tracks used in their work is increased only minimally. The flexibility of the copper foil allows it to bend rather than stretch. This is shown in Figure 3.4-3.

![Figure 3.4-3: Diagram showing the principle of the horseshoe layout. As the substrate is stretched it gets thinner, so that the total conductive path length remains the same.](image)

This allows elongations from 50 to 300 % before the copper ruptured for different reported designs. However, the cyclic stress resulting from repeated stretching cycles caused the designs to rupture at much lower elongations, as shown by the square and triangle plots in Figure 3.4-4, representing two different designs repeatedly elongated to 10, 15 and 20% elongations.
Figure 3.4-4: The number of elongate-relax cycles before failure for two meandering horse-shoe designs at 10, 15 and 20% elongations [52].

This method is not investigated in this thesis because the choice of encapsulating materials is more limited here than in work by Loher et al. Loher used polyurethane lamination to encapsulate their meandering designs while screen printable polyurethanes are used in this thesis. Consequently it is more difficult to create structures with the desired mechanical properties and it was decided that minimising the elongation of the tracks with inelastic textile substrates would be a better method to ensure durability in this thesis.

Neutral axis:

Another important factor is ensuring that stress on the conductive tracks is minimised during bending normal to the plane. By placing the conductive layer correctly the design can take advantage of the neutral axis effect. When a beam structure is bent, one side of the structure is in extension and the other is in compression. The middle of the beam structure, however, is not stressed. The effect is shown in Figure 3.4-5.

Figure 3.4-5: Diagram showing the nature of extension and compression during the bending of a beam.
As previously described, the printed conductive layer in this work will be supported above the textile by an interface layer and enclosed with an encapsulation layer. Both the interface and encapsulation layer materials will be polymer dielectrics. The position of the conductive layer within the polymer dielectric layer structure must be investigated to find positions in which the stress on the conductive layer is minimised. A neutral axis based printed conductive track durability optimisation is described in chapter 6.

3.4.3 Via centralisation

It is necessary for the sensors dispersed around a garment to be connected to central electronics. Portable signal processing and storage electronics are not examined in this thesis to give greater focus to the wearable electrode networks, however centralisation is still considered an important design feature for compatibility with a portable electronics module. In fact, centralisation is useful even with the table-top electronics used in this work as cables can be connected to a central point rather than originating from several points on the body, restricting movement and increasing setup time.

It is logical that the vias through the textile should be fabricated at this central point, as this means that printing is required on only one side of the textile. If vias were fabricated in the middle of a conductive track, one part of the track would have to be printed on one side of the textile and another part on the opposite side of the textile, increasing fabrication complexity. Figure 3.4-6 shows a concept diagram of a network of sensors with centralised vias for connection to external electronics on a T-shirt.

![Figure 3.4-6: Diagram showing the layout of sensors, connection vias and the conductive paths that connect them in a ‘centralised’ network on a T-shirt.](image)

3.5 Designs for functional components

The three functional components in this system are the passive and active skin contact electrodes and the vias. These are stencil printed rather than screen printed to provide a greater thickness more quickly. All three components are fabricated at the terminations of screen printed conductive tracks, with each track having an electrode at one end and a via at the other. This section first describes the screen printed conductive track terminations and then the fabrication of these three functional components.
3.5.1 Screen printed terminations

The terminations of the conductive track are not encapsulated by screen printing, which leaves a conductive surface for the fabrication of functional components. The layer structure at a termination is shown in Figure 3.5-1.

Figure 3.5-1: Concept CAD drawing showing layer structure at terminations.

The fundamental design of a track termination is the same whether it is a via or an electrode. The size of the termination point is often larger for electrodes than connection points, although devices have been fabricated in which the screen printed terminations are identical for either component.

3.5.2 Passive electrode design

The passive electrodes are fabricated by stencil printing directly on to the conductive layer with a waterproof conductive rubber material. This provides a conductive path from the skin surface to the screen printed structure, and in turn to the amplification electronics. This will also protect the silver from abrasion damage and could allow greater biocompatibility than commercial printed pastes. The structure is shown in Figure 3.5-2.

Figure 3.5-2: Concept CAD drawing showing how a screen printed termination is fabricated into a skin contact electrode.
3.5.3 Active electrode design

The active electrode design in this thesis borrows from work by Kang et al [27], using the same circuit design including the selected op-amp and passive components. Kang et al's work has been described in the literature review in section 2.5.2. The emphasis in the design described here is on improving the homogeneity of the fabrication process, by using only stencil and dispenser printing. It is thought that removing the need for the via at the electrode, by having the buffer circuit on the same side of the textile as the electrode, will reduce the complexity of the electrode fabrication compared to Kang et al’s electrode as printing is required on only one side of the textile. Finally, as these electrodes are on rougher, woven textiles than work by Kang, an interface layer will be included in the design.

The design proposed here uses two different stencil prints. The smaller, dielectric stencil is used for encapsulating the components and conductive tracks that make up the buffer amplifier circuit. The thicker stencil uses conductive rubber, and encapsulates the electrode and circuit. It is essential in this design that the circuit is effectively encapsulated, so that there is no unwanted connection to the skin surface and the signal can only travel to the main amplifier via the buffer amplifier circuit. This creates an active electrode with a self-contained circuit. Because the electrodes are less flexible than the conductive paths due to their high thickness, positioning the circuit inside the electrode may improve durability. Figure 3.5-3 shows the screen printed layout (a), the encapsulation of the circuit (b) and finally the full, conductive encapsulation of the circuit and the electrode base (c).

![Diagram of active electrode design]

**Figure 3.5-3**: Active electrode designs. (a) Printed layout and component placement. (b) Encapsulation with dielectric. (c) Encapsulation with conductive layer to complete the electrode.

3.5.4 Via design

The vias must provide an electrical connection between the two sides of the textile. This is achieved using 2-part buttons that are clamped to pierce the screen printed structure and the textile substrate. These buttons make an electrical connection with the screen printed silver on the termination. The corresponding half of the button on the reverse side of the textile can then be used for connecting to electronics.

These connections must be protected from mechanical damage and water ingress. They also must be electrically insulated from the skin on the printed inner surface of the textile. A stencil
printed polymer dielectric encapsulation is used to encapsulate the button and screen printed silver. This will also increase the stiffness of, and thereby reduce the mechanical strain on, the via structure, which will prevent damage to the via. A concept drawing of the via design used in this work is shown in Figure 3.5-4.

![Figure 3.5-4: Concept CAD drawing showing how a screen printed termination is fabricated into a via. Left - separate components of a via. Right - completed via. The textile substrate is not shown.](image)

### 3.6 Conclusions

The designs for the biopotential monitoring garments fabricated in this thesis have been described. Systems will be composed of dry rubber passive or dry rubber active electrodes, with a thickness of 1 to 5 mm and a resistance of less than 1 kΩ, which are connected by conductive tracks to an array of vias. These networks of connections and electrodes will be printed on to a single patch of inelastic textile, which will be either Escalade or Lagonda depending on the outcome of future tests. This printed inelastic textile will be held to the body with one or more strips of an elastic textile called Elasta.

The printed conductive tracks will use a polymer dielectric to create a smooth interface on which thin layers of conductive paste can be printed. These will subsequently be encapsulated by the same dielectric. Each conductive track will connect a skin contact electrode to a via. Design changes to improve the durability and stretchability of these conductive tracks will be investigated further in Chapters 4 and 6.

The passive electrodes will be composed of conductive rubber stencil printed on to patches of silver at the termination of conductive tracks. The active electrodes will be composed of a buffer amplifier encapsulated in a dielectric by stencil printing which has an exposed screen printed patch at its input. Stencil printed conductive rubber is then applied both to enclose the buffer amplifier and to make contact with the exposed stencil printed patch.

Vias will be implemented with 2-part steel buttons which are clamped into the terminations of conductive tracks. Although amplification electronics are not implemented on the garment in this work, the vias for the dispersed electrodes are collected in an array as this methodology would be suitable for future electronics integration on to the textile.
4 - Paste Selection

4.1 Introduction

The selection of materials with suitable electrical and mechanical properties to ensure correct function and durability is crucial for the fabrication of a flexible electronics system. There are also rheology and curing properties, which are important in demonstrating that these materials are suitable for screen and stencil printing on textiles. This chapter describes the selection process for three pastes, which each perform different functions in the previously described designs. The first paste is a screen printable polymer dielectric paste which will be printed and cured to form interface and encapsulation layers for conductive tracks. This is referred to as an interface paste. The second paste is a screen printable conductive paste which is printed and cured to form connection pads for electrodes and vias and conductive tracks between them. The third paste is a stencil printed conductor-filled rubber which is stencil printed and cured to form thick electrodes which provide robust contact stability between the electrodes and the skin.

The fabrication processes for printed electrode networks are described in full in chapter 5. However, because several of the experiments described in this chapter use screen printed samples, the screen printing process must be described. As previously described in the literature review, screen printing involves passing a paste through a patterned mesh screen on to the substrate, resulting in a film with a thickness defined by the emulsion thickness of the screen.

This is performed by passing a polyurethane or rubber squeegee across the screen. Each film resulting from a single print pass is denoted here as a deposit. Several deposits can be passed through the screen, increasing the thickness of the film, before the film is cured by heat or UV light and consequently solidifies. A cured film resulting from one or more deposits is denoted in this thesis as a print. A layer is one or more prints of the same material. For example, the conductive layer in this thesis is fabricated using 1 print, composed of 1 deposit, whereas the interface layer is fabricated using 3-4 prints each composed of 2-6 deposits.

All screen printed structures described in this thesis are fabricated with a DEK 248 semi-automatic screen printer. UV curable pastes are cured in a UV cabinet and heat curable pastes are cured in a box oven. More information on equipment can be found in appendix B.

4.2 Interface paste selection

The properties of the interface paste are important because the conductive path relies on the interface layer for mechanical strength. The thickness of interface for the conductive paths created in this thesis varies, but is usually at least 20 times greater than the thickness of the conductor. Consequently, if the interface paste is prone to tearing or excessive deformation under stress or in wet conditions, the conductive track will be damaged or broken. Other properties are also examined here to ensure the selected material has good compatibility with screen printing.
4.2.1 Interface paste requirements

Basic requirements for the interface paste were determined from previous work. The following list shows the key requirements for the interface paste:

- **Printability**: A viscosity of 1-30 Pa.s, a short curing time and a sufficiently low temperature curing process for compatibility with the selected textiles (<150 °C).
- **Surface energy**: A sufficiently high surface energy for the printing of commercial conductive pastes on to the cured interface surface. This would be achieved by a paste that has a water contact angle of below 60°.
- **Water absorption**: Very low water absorption (<1 % mass increase per hour when fully submerged in 40 °C water.)
- **Tensile strength and flexibility**: A balance of tensile properties is required. A material with insufficient flexibility (a high Young’s modulus) will reduce wearability. However, too much flexibility will allow damaging deformation of the conductive layer. The Young’s modulus should be between 0.02 and 0.2 MPa.

There are several UV-curable polyurethane pastes that have been developed by the University of Southampton specifically for this application. These polyurethane interface pastes have several useful properties. The UV curing mechanism speeds up fabrication compared to heat curing, which is especially useful as several layers are required to create a smooth surface on a textile for the printing of conductive pastes.

The polyurethane pastes are composed of four main ingredients. These are a polyurethane base, a thinner, a cross-linking agent and a UV-activated initiator for the cross-linking process. The four pastes used here contain various amount of each of these, although the exact chemical composition cannot be disclosed. The measured viscosities and curing times of the materials examined in this section are given in Table 4-1.

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Curing time (s)*</td>
<td>20</td>
<td>60</td>
<td>40</td>
<td>20</td>
</tr>
<tr>
<td>Measured Viscosity (Pa.s)</td>
<td>6.2</td>
<td>3.6</td>
<td>11.1</td>
<td>15.5</td>
</tr>
</tbody>
</table>

* with a curing power density of 31 mW/cm² at 365nm.

The viscosity was measured using a Brookfield CAP1000+ cone and plate viscometer. This is referred to as the standard viscosity measurement apparatus and the viscosity measuring process is described in greater detail in appendix B.

Each of these pastes is translucent, so UV light that is not reflected can pass through the material during curing. If the materials were not translucent only the surface would be cured. The curing times given are approximate because the curing time depends not only on the curing energy provided by the UV light source but also on the level of internal reflection and light absorption by the textile.

4.2.2 Interface paste printability
The interface pastes were each printed on to Escalade textile to find the necessary number of deposits to create a smooth surface. The Escalade textile shown in Figure 4.2-1 has been printed with four layers of UoS-IF-#4. The screen design used is shown in Figure 4.2-2. The conductive layer screen design is shown in red and the interface layer screen design is shown in green.

The first print for each paste consists of six deposits as this is necessary for the polyurethane to sink into the textile and bond firmly, creating a non-porous foundation for subsequent prints. The next prints have a number of deposits defined by the roughness of the surface. A print with more deposits tends to be more effective in smoothing the surface quickly but also uses more paste. The aim is to use as little paste as possible, which will minimise the layer thickness and speed up the fabrication process. The separate prints of interface paste bond firmly to one another during curing.

Table 4-2 shows the number of deposits required to fabricate a smooth layer with each polyurethane interface paste. The fabrication time is calculated using the curing times shown in Table 4-1.

**Table 4-2: Print processes to create a smooth interface with each polyurethane interface material.**

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Print 1</td>
<td>6 Deposits</td>
<td>6 Deposits</td>
<td>6 Deposits</td>
<td>6 Deposits</td>
</tr>
<tr>
<td>Print 2</td>
<td>4 Deposits</td>
<td>4 Deposits</td>
<td>4 Deposits</td>
<td>4 Deposits</td>
</tr>
<tr>
<td>Print 3</td>
<td>2 Deposits</td>
<td>2 Deposits</td>
<td>2 Deposits</td>
<td>2 Deposits</td>
</tr>
<tr>
<td>Print 4</td>
<td>1 Deposit</td>
<td>2 Deposits</td>
<td>1 Deposit</td>
<td>1 Deposit</td>
</tr>
<tr>
<td>Print 5</td>
<td>-</td>
<td>2 Deposits</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Print 6</td>
<td>-</td>
<td>2 Deposits</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Fabrication Time* (minutes)</td>
<td>4.17</td>
<td>10</td>
<td>6.83</td>
<td>5.5</td>
</tr>
</tbody>
</table>

*Assuming a print time of 10 seconds per deposit and a 5 second movement period between printing and curing.
The pastes UoS-IF-#4, UoS-IF-#39 and UoS-IF-#44 took the same number of deposits to appear smooth to the naked eye, although the fabrication time varied due to their different required curing conditions. UoS-IF-#35 took significantly more prints to create a visibly smooth surface, and consequently had a longer fabrication time than the other three.

A series of photographs and scanning electron microscope (SEM) micrographs showing the fabrication of a smooth interface of UoS-IF-#4 on Escalade textile are shown in Figure 4.2-3, Figure 4.2-4 and Figure 4.2-5. All micrographs in this thesis are taken with a Zeiss EVO SEM. Figure 4.2-3 shows cross sectional SEM micrographs of the printed structure and textile substrate as the number of prints increases from 1 to 4.

![Figure 4.2-3: SEM micrographs of the polyurethane print process as the number of prints increases from 1 to 4 - cross sectional view.](image)

Although there have never been any issues with adhesion between layers of polyurethane, the lines marking the divisions between separately cured prints are visible on these micrographs. The micrographs in Figure 4.2-4 shows the change in the polyurethane surface as the number of prints increases from 1 to 4.

![Figure 4.2-4: SEM micrographs of a printed polyurethane surface on textile as the number of prints increases from 1 to 4 - top-down view.](image)

The surface gets smoother as more prints are added. Pinholes, however, can still be observed in the surface after 4 prints. A pinhole with a diameter of around 50 µm can be observed in the top-left of the third micrograph in Figure 4.2-4. The origin of these pinholes is unclear but they are too small and appear too infrequently to dramatically affect the measured resistance. They may be caused by non-uniform parts of the textile which are more porous than the surrounding region, or by air bubbles in the paste that shift during printing or curing.
Figure 4.2-5 shows the change in surface structure as observed with the naked eye. It is problematic to use a stylus based profilometer, the only one available for this work, on a textile so a ‘smooth’ interface is defined here as one that appears smooth to the naked eye.

Further discrimination between different levels of smoothness can be determined after a conductive layer is printed on the surface of the interface. A high resistance corresponds to a heterogeneous conductor cross section caused by a rough interface surface, whereas a low resistance indicates a homogeneous conductor cross section caused by a smooth interface surface.

To investigate, a single layer of Dupont 5000 silver conductive paste was printed on to cured interfaces fabricated with each of the four examined polyurethane pastes. The interfaces were printed and cured as described on page 57. Screen PC-1 was used.

After printing the Dupont 5000, the samples were cured for 10 minutes at 120 °C producing 5-10 µm thick conductive tracks. The resistance of each track was then measured with a Tenma 72-7735 digital multimeter. All resistance measurements in this thesis are taken with this multimeter unless indicated otherwise. Point to point measurements, rather than four-point measurements which require a conductive square to be printed, were taken to allow more measurements to be taken from each printed tile with greater speed and minimise the use of silver pastes.

The sheet resistivity of the cured conductive tracks and the corresponding standard deviations are shown in Table 4-3. Each measurement is an average of 16 printed conductive tracks of lengths 20 to 50 mm and width 1 mm.

<table>
<thead>
<tr>
<th></th>
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<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Sheet resistivity (Ω/square)</td>
<td>0.09</td>
<td>0.21</td>
<td>0.14</td>
<td>0.17</td>
</tr>
<tr>
<td>Standard deviation (Ω/square)</td>
<td>0.01</td>
<td>0.10</td>
<td>0.04</td>
<td>0.03</td>
</tr>
</tbody>
</table>

These measurements indicate that although the surfaces of UoS-IF-#4, UoS-IF-#39 and UoS-IF-#44 appeared smooth to the naked eye, the differences between them are more pronounced in the effect on the resistivity of a printed conductive track. These results suggest UoS-IF-#4
can be used to create a smoother interface surface than the other pastes, using the same number of interface deposits and prints.

To conclude, UoS-IF-#4 can be used to create a very smooth surface in a short space of time. UoS-IF-#39 and UoS-IF-#44 are difficult to separate in terms of printability, with UoS-IF-#39 having a slightly smoother surface after four layers but a slightly longer curing time. UoS-IF-#35 cures slowly and is difficult to use to create a smooth surface on the textile.

4.2.3 Surface energy of interface material

The surface energy is an important property because it dictates how well the surface of the cured material wets and how well the next printed layer will bond to it. A comparison between two interface layers composed of UoS-IF-#4 and N294 silicone rubber shows clearly the effect of surface energy when the two surfaces are printed with a thinned formulation composed of Dupont 5000 silver and T-402 thinner [65]. This thinned formulation was used specifically to exaggerate the effects of differences in substrate surface energy, which can be observed more clearly when the viscosity of the deposited substance is low. Figure 4.2-6 shows that the paste spreads evenly over the polyurethane interface on the left, but agglomerates into droplets on the silicone rubber on the right.

![Figure 4.2-6: A thinned silver paste printed on to a high surface energy polyurethane (left) and a low surface energy silicone rubber (right).](image)

Vulcanised silicone rubber has very few free chemical bonds to attach to a subsequent layer and consequently adhesion as well as surface energy will be low. This affects durability through poor adhesion as well as printability through low surface energy. Factors such as mechanical bonding also have a significant effect on the level of adhesion.

The effect of strong adhesion between a thin conductive layer and a substrate is that the conductor follows any deformation of the substrate. The conductive layer is composed of a matrix of conductive particles suspended in a polymer. If the substrate is flexible, extension will cause the conductivity of the conductive layer to decrease as these conductive particles are moved apart, but when the substrate returns to its original shape the particles are likely to return to similar positions and conductivity will return to the same level. If the adhesion is low
there is more reliance on the mechanical properties of the conductive layer itself during deformation. A concept diagram showing the effect of low conductor-substrate adhesion during bending is shown in Figure 4.2-7.

![Figure 4.2-7: Difference in conductive track durability on a flexible substrate with high surface energy and good adhesion (a), and low surface energy and poor adhesion (b).](image)

The surface energies of the polyurethanes are examined in this section with a contact angle measurement and with a scratch test.

### 4.2.3.1 Contact angle measurement on cured polyurethane layers

A contact angle measurement uses a liquid with a known surface tension to estimate the surface energy of a given material. A drop of the liquid is placed via a syringe on to the material surface. The drop will form a droplet shape on the material surface. The shape of this droplet is dependent on the surface energy of the material. The shape is described by the contact angle, the angle that the droplet wall makes against the material surface. In this work a Kruss DSA30 tensiometer is used to perform the drop test. The drop deposition process is shown in Figure 4.2-8.

![Figure 4.2-8: Deposition of a drop of DI water on to a printed polyurethane surface.](image)

A high substrate surface energy will exceed the surface tension of the liquid and the drop will spread out and flatten, leading to a low contact angle. A lower substrate surface energy will allow the liquid to pull together and form a rounder droplet, giving a high contact angle. The four polyurethane materials are examined using drops of deionised (DI) water. Four repeats, using a new sample each time from the same printed batch, were performed for each material. The contact angle was measured after 5 and 30 seconds and the results presented here are an average of the two measurements. The volume of the drop was between 1.5 and 4 µL in all tests. The contact angle is measured automatically by the equipment. The average contact angles of the four materials are shown in Table 4-4.
Table 4-4: Contact angle for deionised water on various polyurethanes

<table>
<thead>
<tr>
<th></th>
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<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Contact Angle</td>
<td>44.8</td>
<td>94.4</td>
<td>102.4</td>
<td>100.5</td>
</tr>
<tr>
<td>Standard Deviation</td>
<td>5.9</td>
<td>1.6</td>
<td>2.6</td>
<td>4.3</td>
</tr>
</tbody>
</table>

These results show that UoS-IF-#4 has a lower contact angle, indicating that it has a significantly higher surface energy than the other polyurethanes. UoS-IF-#35 has a lower surface energy than UoS-IF-#4 but slightly higher than UoS-IF-#39 and UoS-IF-#44. Some knowledge of the chronology of the development of the pastes is useful to understand these results. UoS-IF-#4 was developed before the other pastes, with minimal consideration given to the water absorption. After this paste was developed, additives were introduced to newly developed pastes to improve water resistance. These additives reduced the surface energy of cured layers fabricated using these pastes. The water absorption of polyurethanes is discussed in more detail later in this chapter.

4.2.3.2 Abrasion resistance of a conductive layer on polyurethane

To give a practical measure of the surface energy of the printed interface layers, a scratch test was performed. Samples with interface layers composed of each polyurethane material were printed with a single deposit of Dupont 5000. Screen design PC-1 was used. These were cured for ten minutes at 100 °C. The curing temperature for Dupont 5000 is 120 °C but partially curing reduces the adhesion so that a scratch test causes a measurable resistance change on all polyurethane interface materials. These samples were then heated at 80 °C for 10 minutes and peeled from the alumina tile, which is the removal process for all printed textiles in this thesis.

The resistance of each track was measured. The textile was then peeled from the alumina tile used to keep it flat during printing then cut into patches of four. Eight conductive tracks, in two patches of four, are tested for each interface material. The printed samples were subjected to abrasion by scratching the printed conductive surface for 15 seconds with a spatula. The spatula is scraped perpendicular to the conductive tracks by hand. The resistance of each conductive track after scratching is measured. Figure 4.2-9 shows the increase in conductive track resistance after the scratch test.
Figure 4.2-9: Results of scratch tests performed on a Dupont 5000 conductive track printed on interfaces composed of different polyurethane pastes.

It is clear from this test that the adhesion is significantly worse between the conductive paste and the UoS-IF-#39 interface paste. Given that this paste had roughly the same water contact angle as UoS-IF-#44, there may be factors other than surface energy, such as tensile properties, that also affect this result. UoS-IF-#4, UoS-IF-#35 and UoS-IF-#44 all have smaller increases in resistance after the test, suggesting comparatively better adhesion and higher surface energy. The lowest resistance change after abrasion occurred on the UoS-IF-#4 interface paste.

The manual method of abrading the conductive tracks will not have provided a stable scratching pressure for all samples. However, effort was made to treat each tested sample the same and the standard deviation was sufficiently low to show clearly the best and worst interface surfaces in terms of adhesion. These results correlate with the lowest and highest contact angles respectively, as measured in the previous section.

4.2.4 Water absorption properties of the interface materials

Water absorption is an important property of the interface layer because one of the objectives is for these systems to be washable. If the material absorbs water it will cause dimensional changes in wet conditions that could be damaging to the conductive layer and could degrade the mechanical strength of the interface.

The amount of water absorbed by a given interface material is found by weighing a sample of the material before and after a period of submersion in water in a beaker placed on a hot plate. Figure 4.2-10 shows the water absorption of UoS-IF-#4 in 40 °C DI water over 60 minutes.
Each point on the graph corresponds to the absorption of a stencil printed sample of dimensions 30 x 30 x 1.2 mm submerged for the indicated period. After the pre-defined submersion period the sample surface is dried with a lint-free wipe before weighing. The scales used are accurate to +/-0.01 grams, equivalent to 0.66% with a typical sample weight of roughly 1.5 grams.

The same stencil was used to fabricate square samples of each polyurethane material. Four samples are used for each material. The submersion period is 24 hours and the water temperature is 40 °C. DI water is used. Table 4-5 shows the weight change for each material.

<table>
<thead>
<tr>
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<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Absorption (%wt/day)</td>
<td>21.1</td>
<td>0</td>
<td>0.7</td>
<td>0</td>
</tr>
</tbody>
</table>

This experiment showed that UoS-IF-#4 had far higher water absorption than the other pastes. UoS-IF-#35 and UoS-IF-#44 showed no measurable water absorption, while the absorption of UoS-IF-#39 is low enough to be negligible during a couple of hours of submersion as might be encountered during a typical domestic wash. However, this small level of water absorption may affect the properties of the material over a long period.

### 4.2.5 Tensile properties of interface material

The tensile properties of the cured polyurethane will affect the durability of the device. If the level of elongation before break is too low the conductive tracks will break easily during planar extension of the textile or during folding and bending. The standard tensile test apparatus is used to tensile test stencil printed polyurethane samples to failure. Dumbbell shaped samples are used in this thesis for the tensile testing of polymers, with gripping surfaces wider than the gauge section to ensure the ends of samples cannot slip out of the grips of the tensile test machine when they are deformed. The dumbbell stencil has a thickness of 1.2 mm and a gauge length and width of 20 mm and 10 mm respectively. The gripping surfaces have a width of 30 mm. The dumbbell stencil design is shown in Figure 4.2-11.
A tensile test of each of the four polyurethane materials was carried out using the standard tensile apparatus with an extension speed of 200 mm/minute. Four samples were tested for each material and the average is taken from the four.

The standard deviations for each property extracted from the results for each material are provided later in Table 4-6. An example showing graphically the typical variation in tensile properties between tensile samples of the same material is provided in appendix B. The variation is low enough to give confidence that the averaged results accurately represent the tensile properties of the material. The averaged test results, and crosses showing the point of failure of each individual sample, are shown in Figure 4.2-12.

**Figure 4.2-11**: A diagram of the stencil printed dumbbell samples used for all the tensile testing of polymers and polymer composites in this thesis.

**Figure 4.2-12**: Tensile tests of stencil printed samples of the four polyurethanes. Each measurement is the average of four tensile tests.

The important properties to extract from these results are the elongation at break, which must be high to allow bending of the printed track without rupture, the tensile strength, which indicates how much stress will be required to break the printed track, and the Young’s modulus, which describes the relationship between stress and strain for a material during elastic deformation.

**Table 4-6** shows the calculated properties and the standard deviation for the four polyurethanes tested. The Young’s modulus is calculated using the stress and extension at
break, although the actual modulus, given by the gradient of the curve, varies over the course of the test.

Table 4-6: Summary of the tensile properties of the four polyurethanes.

<table>
<thead>
<tr>
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<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Tensile strength (MPa)</td>
<td></td>
<td>7.59</td>
<td>0.98</td>
<td>1.30</td>
<td>3.94</td>
</tr>
<tr>
<td>Standard deviation</td>
<td></td>
<td>1.16</td>
<td>0.15</td>
<td>0.22</td>
<td>0.62</td>
</tr>
<tr>
<td>Elongation at break (%)</td>
<td></td>
<td>33.24</td>
<td>13.54</td>
<td>84.80</td>
<td>25.28</td>
</tr>
<tr>
<td>Standard deviation</td>
<td></td>
<td>4.67</td>
<td>3.28</td>
<td>5.72</td>
<td>3.99</td>
</tr>
<tr>
<td>Young’s modulus (MPa)</td>
<td></td>
<td>0.228</td>
<td>0.072</td>
<td>0.015</td>
<td>0.156</td>
</tr>
<tr>
<td>Standard deviation</td>
<td></td>
<td>0.0170</td>
<td>0.0082</td>
<td>0.0019</td>
<td>0.0058</td>
</tr>
</tbody>
</table>

UoS-IF-#39 has the best tensile properties for high flexibility, with a very low Young’s modulus and a high elongation at break. It also has a relatively constant Young’s modulus over a range of extensions, indicating that its deformation may be elastic rather than plastic and it is consequently more likely to return to its original shape when an applied stress is removed. UoS-IF-#4 is the strongest and stiffest material, with the highest tensile strength and highest Young’s modulus. It also has the second highest elongation at break. UoS-IF-#35 and UoS-IF-#44 have comparatively poor tensile properties.

A further tensile test was carried out to illustrate the effect of water ingress into UoS-IF-#4 on its tensile properties. A water ingress cycle was defined as 24 hours in 40 °C deionised water, followed by a drying period of 90 minutes, also at 40 °C. Sets of dumbbell samples of UoS-IF-#4 with a thickness of 1.2 mm and a gauge length and width of 20 mm and 10 mm respectively were subjected to one and three of these water ingress cycles. These samples, along with samples that hadn’t been submerged, were tensile tested with the standard tensile test apparatus. Four samples were used for each level of water ingress (0, 1 and 2 cycles) and their tensile properties and individual break points are shown in Figure 4.2-13.

![Figure 4.2-13: Tensile test of dumbbell samples of UoS-IF-#4 after varying periods of exposure to wet conditions.](image)

This test shows that water ingress cycles do not dramatically alter the Young’s Modulus but do reduce the elongation at break. This is believed to be due to microscopic fractures formed as the water moves in and out of the material. This shows that UoS-IF-#4, if used in a conductive
track structure, should not be exposed to moisture as it will absorb this moisture and the result will be a degradation of its tensile properties.

4.2.6 Selected interface material

Table 4-7 shows a summary of the interface material properties.

<table>
<thead>
<tr>
<th></th>
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<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Curing time (s)*</td>
<td>20</td>
<td>60</td>
<td>40</td>
<td>20</td>
</tr>
<tr>
<td>Viscosity (Pa.s)</td>
<td>6.2</td>
<td>3.6</td>
<td>11.1</td>
<td>15.5</td>
</tr>
<tr>
<td>Prints (Deposits) required</td>
<td>4(13)</td>
<td>6(18)</td>
<td>4(13)</td>
<td>4(13)</td>
</tr>
<tr>
<td>Contact angle</td>
<td>44.8</td>
<td>94.4</td>
<td>102.4</td>
<td>100.5</td>
</tr>
<tr>
<td>Scratch test (% increase)</td>
<td>16.3</td>
<td>13.2</td>
<td>123.2</td>
<td>36.6</td>
</tr>
<tr>
<td>Young’s Modulus</td>
<td>0.228</td>
<td>0.072</td>
<td>0.015</td>
<td>0.156</td>
</tr>
<tr>
<td>Elongation at break (%)</td>
<td>33.24</td>
<td>13.54</td>
<td>84.80</td>
<td>25.28</td>
</tr>
<tr>
<td>Water absorption (%/day)</td>
<td>21.1</td>
<td>0</td>
<td>0.7</td>
<td>0</td>
</tr>
</tbody>
</table>

*at curing power density 31mW/cm² at 365nm.

These measured values can be used to select the paste which is most appropriate for use as an interface and encapsulation material.

Firstly, UoS-IF-#35 has too low a viscosity to print properly and provide a smooth surface, while the other pastes can be printed easily. UoS-IF-#39 and UoS-IF-#44 take slightly longer to cure, however they are still fast compared to most heat curing pastes which typically have curing times from 3-30 minutes. UoS-IF-#4 has very high water absorption, making it unsuitable for use in a washable device. The other interface pastes have very low water absorption.

UoS-IF-#39 has low surface energy, and consequently printed pastes adhere poorly, while the other pastes have higher surface energy and print easily. UoS-IF-#44 and especially UoS-IF-#35 have comparatively poor tensile properties. Given that durability of the tracks is very important, UoS-IF-#44 and UoS-IF-#35 are ruled out from use in this system. This leaves UoS-IF-#4 and UoS-IF-#39 as suitable pastes. These pastes could be used independently, however given that UoS-IF-#4 has high surface energy and UoS-IF-#39 has negligible water absorption, it is possible to stack separate layers of the materials so that an outer layer of UoS-IF-#39 prevents water absorption and an inner layer of UoS-IF-#4 has good adhesion to the conductive layer, as described by Yang et al [66]. This design is shown in Figure 4.2-14.

![Figure 4.2-14: Proposed layout for polyurethane interface and encapsulations.](image-url)
Although the optimal pastes have been identified, further work is required to fully define the optimal interface and encapsulation layer structures. Experiments are performed with various layer structures composed of UoS-IF-#4 and UoS-IF-#39 in Chapter 5.

4.3 Conductive paste selection

There are several commercially available conductive pastes used for flexible electronics. This section defines the requirements of the conductive paste and examines several candidate pastes.

4.3.1 Conductive paste requirements

The following list shows the key requirements for the conductive paste. These requirements are determined from the literature:

- **Printability**: A viscosity of 1-30 Pa.s, a short curing time and a sufficiently low temperature curing process for compatibility with the selected textiles (>150 °C).
- **Resistivity**: A low resistivity, to provide conductive tracks of 100 cm, a length sufficient for most torso applications, with a resistance under 100 Ω. 100 Ω is chosen based on the electrode-skin resistance levels typically encountered of 1-1000 kΩ, a 100 Ω track will not add significantly to these. If it is assumed that the track width is 2 mm, the resistivity of a single layer must be under 0.2 Ω/square.
- **Adhesion**: The silver paste should have good adhesion to the UoS-IF-#4 polyurethane interface layer.
- **Flexibility**: The paste should have high flexibility, giving a low resistance change (<50% increase) after expected levels of deformation such as a machine wash.

Silver polymer pastes are examined in this work because many of these pastes meet these requirements. These have high conductivity and good flexibility. The curing temperatures of the pastes selected are all less than 150 °C, to ensure compatibility with textiles. The pastes are summarised in Table 4-8. The viscosity is measured with the standard viscosity measurement apparatus.

**Table 4-8: Curing time and measured viscosity for the examined silver pastes.**

<table>
<thead>
<tr>
<th>Paste</th>
<th>Dupont 5000</th>
<th>Conductive Compounds AG-800</th>
<th>Electrodog 725A</th>
<th>UoS-TC-32</th>
</tr>
</thead>
<tbody>
<tr>
<td>Curing time (minutes)</td>
<td>10</td>
<td>1.5-6</td>
<td>15</td>
<td>10</td>
</tr>
<tr>
<td>Curing temperature (°C)</td>
<td>120</td>
<td>130</td>
<td>120</td>
<td>120</td>
</tr>
<tr>
<td>Silver content (%)</td>
<td>58-62</td>
<td>54-58</td>
<td>58-62</td>
<td>60</td>
</tr>
<tr>
<td>Datasheet Viscosity (Pa.s)</td>
<td>3.5-16</td>
<td>12</td>
<td>11-14</td>
<td>Not given</td>
</tr>
<tr>
<td>Measured Viscosity (Pa.s)</td>
<td>3.742</td>
<td>2.741</td>
<td>5.419</td>
<td>1.567</td>
</tr>
</tbody>
</table>

*UV curing

4.3.2 Resistivity of silver pastes

To test the silver pastes a different set of screens was used so that all printed conductive tracks were the same length. This screen design also implemented gradual curves at the points where the conductive tracks meet termination pads, where the previous screen design (PC-1)
had internal corners. This was implemented to improve durability, as internal corners tend to be points at which stress accumulates in a thick film structure. The screen design is shown in Figure 4.3-1 and a sample of Lagonda textile printed with this screen is shown in Figure 4.3-2.

![Figure 4.3-1: Second screen design PC-2 interface (green) and PC-2 conductor (red).](image1)

![Figure 4.3-2: Printed conductive tracks using screen designs PC-2, UoS-IF-#4 interface and UoS-TC-32 conductor.](image2)

In order to gauge the resistivity of the silver pastes each paste is printed in a single deposit on to a textile printed with an interface layer of UoS-IF-#4. These are then cured at the appropriate temperature as defined in the materials’ datasheets and shown in Table 4-8 above. There are twelve printed tracks of length 40 mm on each tile. Six of these have a width of 1 mm and the other six have a width of 2 mm. The resistance of each conductive track is measured. The average sheet resistivity and the standard deviation from the average for each paste are shown in Table 4-9.

<table>
<thead>
<tr>
<th>Paste</th>
<th>Dupont 5000</th>
<th>Electroag 725A</th>
<th>Conductive Compounds Ag-800</th>
<th>UoS-TC-32</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sheet resistivity (Ω/square)</td>
<td>0.086</td>
<td>0.056</td>
<td>0.051</td>
<td>0.069</td>
</tr>
<tr>
<td>Standard deviation (Ω/square)</td>
<td>0.0054</td>
<td>0.0025</td>
<td>0.0079</td>
<td>0.0040</td>
</tr>
</tbody>
</table>

Electroag 725A and Conductive compounds Ag-800 have the lowest resistivity. Dupont 5000 has the highest resistivity. The resistivity of every conductive paste tested is below 0.1 Ω/sq and significantly under the target maximum resistivity (0.2 Ω/sq).

### 4.3.3 Resistance to abrasion of silver pastes

The resistance to abrasion is used here to gauge the adhesion of the silver pastes to a polyurethane substrate. If the adhesion is poor, scratching the surface of the silver track will cause parts of the silver track to be removed from the substrate, which will increase the measured resistance of the remaining conductive track.
Four samples of Lagonda textile were printed using screen design PC-2 to create an interface layer of UoS-IF-#44 with a thickness of 100 µm. This provided a smooth surface on the textile. UoS-IF-#44 is used because it performed worse than UoS-IF-#4 in the scratch test described in section 4.2.3.2 and therefore differences in abrasion resistance between silver pastes printed on to its surface will become apparent more quickly.

Each tile was then printed with a single deposit of one of the silver pastes examined in this section, giving a single printed textile sample for each silver paste. These were cured as before. Each textile sample provided six 40 x 1 mm silver tracks on polyurethane interfaces. The resistance of each printed track was measured.

These tracks were subjected to a 15 second scratch test using a spatula, scratching perpendicular to the conductive tracks, and the resistance was measured again. Each test used a patch with three tracks, as shown in Figure 4.3-3. Two 3-track patches were tested for each silver paste so each result is an average of six tested conductive tracks. The percentage increases in resistance after the scratch test for printed tracks composed of each silver paste are shown in Figure 4.3-4.

These results show that UoS-TC-32 is significantly less durable to abrasion compared to the commercial pastes. The Dupont, Electrodag and Conductive Compounds silver pastes all have comparatively good abrasion durability when cured, although Electrodag 725A has a slightly higher resistance increase than Dupont 5000 or Conductive Compounds Ag-800.

As with the previous scratch test, the printed conductive tracks were scratched manually and there was no way of ensuring that the force was constant. However, this test showed a significant difference between the behaviour of the commercial and UoS conductive materials which is used to support other observations in this section.

4.3.4 Flexibility of silver pastes

The flexibility of the examined silver pastes is tested here with a machine mandrel test. This test involves repeatedly passing a printed sample around a given bending radius cylinder, also known as a mandrel, and gauging the effect on durability by periodically removing printed samples from the machine and taking resistance measurements. Although this test can be performed by hand, a machine mandrel is preferable to reduce test time, increase the number of bending cycles a sample can be subjected to and improve the repeatability of the
deformations. The mandrel apparatus used in this work has two pockets on the mandrel belt into which samples can be inserted, so that two conductive tracks can be tested simultaneously. The belt itself has a width of 50 mm. The machine mandrel is shown in Figure 4.3-5. More information on this equipment is found in appendix B.

![Image of the machine mandrel](image)

**Figure 4.3-5:** The machine mandrel, which is used to test the effect of cyclic bending on a printed track.

A UoS-IF-#4 interface is printed on to Escalade textile in a layer with thickness 150 µm using the PC-2 screens described in the previous section. This interface creates a smooth surface on the textile. Each of these samples is then printed with a single deposit of a different silver paste. The length of the silver track is 40 mm and the width is 1 mm, and the polyurethane interface and encapsulation width is 2 mm. The samples are then encapsulated with a UoS-IF-#4 layer of 2 deposits which has a thickness of around 60 µm. A thicker encapsulation would provide greater durability, but a thinner encapsulation is used in this test to allow differences in durability to become apparent more quickly.

The printed tracks are subjected to a mandrel test. This is performed with the printed structure facing in to the bending radius and the textile on the outside, a regime denoted in this thesis as internal bending. Four conductive tracks are tested for each type of silver paste. Each conductive track is subjected to 200 mandrel cycles and removed for a resistance measurement after 1, 2, 3, 5, 10, 20, 50, 80, 130 and 200 cycles. The results for a test with 200 mandrel cycles at radius 3mm is shown in Figure 4.3-6. Error bars show 1 standard deviation.
The resistance is shown here normalised with reference to the initial resistance \( R_0 \) so that this test focuses entirely on the behaviour of the paste after bending cycles. The initial resistance of each paste has already been examined. This test shows clearly the different response to deformation of the silver polymer pastes tested. This can be thought of as an examination of the elasticity of the cured pastes, as the volume of silver in the conductive track does not change, it is merely displaced. If the particles return to approximately their initial positions after the track is bent, the change in resistivity will be low and the elasticity of the polymer paste is high. If the deformation is more plastic and less elastic the distribution of the particles will change after the deformation and there will be a greater increase in resistivity.

The mandrel process was very repeatable and the low level of deviation between samples of the same conductive material allows Electrodag 725A to be distinguished as the most elastic conductive paste, despite Conductive Compounds Ag-800 also performing relatively well. As in the scratch test, UoS-TC-32 shows very poor durability.

### 4.3.5 Selected silver paste

Table 4-10 shows a summary of the properties of the examined silver pastes.

<table>
<thead>
<tr>
<th>Paste</th>
<th>Dupont 5000</th>
<th>Electrodag 725A</th>
<th>Conductive Compounds Ag-800</th>
<th>UoS-TC-32</th>
</tr>
</thead>
<tbody>
<tr>
<td>Curing temperature (°C)</td>
<td>120</td>
<td>120</td>
<td>130</td>
<td>120</td>
</tr>
<tr>
<td>Viscosity (Pa.s)</td>
<td>3.742</td>
<td>5.419</td>
<td>2.741</td>
<td>1.567</td>
</tr>
<tr>
<td>Resistivity (Ω/Squ)</td>
<td>0.086</td>
<td>0.056</td>
<td>0.051</td>
<td>0.069</td>
</tr>
<tr>
<td>Scratch test (% increase)</td>
<td>1.2258</td>
<td>1.3304</td>
<td>1.2076</td>
<td>9.1865</td>
</tr>
<tr>
<td>Mandrel (% increase after 200 cycles)</td>
<td>6.43</td>
<td>3.58</td>
<td>4.27</td>
<td>30.97</td>
</tr>
</tbody>
</table>

Electrodag 725A is the best conductive layer material as it has the lowest sheet resistivity as well as the best resistance to abrasion and damage during flexure. Conductive Compounds Ag-
800 also performed well on all of these tests, although the low standard deviation observed in the mandrel test allows a distinction to be made between these two more durable conductive materials. UoS-TC-32 is the poorest performing silver paste examined and has significantly worse durability than the other three.

4.4 Conductor loaded rubber paste selection

The conductor loaded rubber paste must be conductive, but has different requirements from the conductive paste described above. A higher resistivity is acceptable but the material must be more durable because unlike the conductive track material, the electrodes are exposed. It is also printed in a thickness hundreds of times greater than the thickness of the conductive track and so must be a cheaper material.

Unlike the interface and conductive pastes, this paste was formulated as part of this project and is one of the novelty claims of this thesis. Consequently this section contains a more detailed examination of the chemical properties and production of this material. The target requirements for this paste are based on electrodes examined in the literature review and designs described in the previous chapter.

4.4.1 Conductor loaded rubber paste requirements

- **Printability:** A stencil printable viscosity of 20-100 Pa.s, a short curing time and a sufficiently low temperature curing process to be compatible with the textiles used in this thesis (<150 °C).

- **Resistivity:** A low resistivity, to provide electrodes of thickness up to 3 mm with resistance under 200 Ω. If the minimum electrode diameter is assumed to be 10 mm, the resistivity of the material should be less than 520 Ω.cm. This target is relatively arbitrary, and is simply based on the fact that lower resistance is better for reducing impedance mismatch between electrodes. The influence of rubber electrode conductivity on performance is described later in section 7.3.3.

- **Durability:** The material must have sufficient tensile strength to withstand impacts during machine washing. Some elasticity is also required to increase device flexibility and user comfort. The material must not absorb or react with water.

- **Cost:** Since the layer thickness for the conductor loaded rubber (3 mm) is around 300 times that of the conductive layer (10 µm), a comparatively large amount of this material is used and the cost of this material should be significantly less than that of silver polymer pastes.

Given these requirements, carbon loaded silicone rubber (hereafter referred to as carbon loaded rubber) was selected as a conductive material for electrodes. This has higher resistivity than a silver polymer paste but is much cheaper, waterproof and can be formulated into a stencil printable paste. Conductor loaded silicone rubbers have already been used in several skin contact medical applications and biocompatible, medical grade silicone rubbers are available from the large silicone rubber producers, such as Wacker and Dow Corning. The silicone rubbers selected are two part room temperature vulcanising (RTV) silicone rubbers. These begin to vulcanise when mixed together and this vulcanisation can be accelerated by
increasing the temperature. By mixing in carbon black before vulcanisation, the material can be made conductive. Table 4-11 shows the properties of the four commercially available silicone rubbers examined in this work.

<table>
<thead>
<tr>
<th>Paste</th>
<th>SilGel 612</th>
<th>Sylgard 184</th>
<th>Viscolo 22</th>
<th>RTV-M</th>
</tr>
</thead>
<tbody>
<tr>
<td>Manufacturer</td>
<td>Wacker</td>
<td>Dow Corning</td>
<td>TOMPS</td>
<td>Dow Corning</td>
</tr>
<tr>
<td>Curing time (minutes)</td>
<td>15</td>
<td>30</td>
<td>20</td>
<td>20</td>
</tr>
<tr>
<td>Curing temperature (°C)</td>
<td>100</td>
<td>100</td>
<td>100</td>
<td>80</td>
</tr>
<tr>
<td>Viscosity (Pa.s)</td>
<td>1</td>
<td>3.9</td>
<td>4</td>
<td>90</td>
</tr>
</tbody>
</table>

In this work ENSACO 250G carbon black [67] is used as the carbon material. Two solvent thinners, ESL T-402 [68] and Dow Corning SL9106 [69], are also trialled for the purpose of reducing viscosity and slowing down vulcanisation in order to ease the stencil printing process. T-402 is a solvent thinner based on butyl carbitol acetate and SL9106 contains terminated siloxane molecules which inhibit vulcanisation and can be used as a silicone release agent or a thinner. Preliminary work was carried out with Dow Corning RTV-M and then the other silicone rubbers were investigated to reduce the viscosity and improve mixing. Some work in this section refers to screen printed samples, which were investigated at the start of this project. This was not continued as stencil printing had a wider range of acceptable viscosities and was more appropriate for fabricating layers of the required thickness.

4.4.2 Production of carbon loaded rubber

Silicone rubbers are used in many applications, one of which is as a dielectric to encapsulate electronics. The dielectric strength of the silicone rubbers used in this work is around 20 MV/m. As previously mentioned, adding carbon black to a silicone rubber before vulcanisation will make the material conductive. The more carbon that is added to the mixture, the higher the conductivity of the vulcanised material. At low levels of carbon loading the carbon particles are physically isolated from one another and the conductivity remains low. When sufficient carbon has been added to create conductive paths through the material the conductivity increases sharply. The level of carbon loading at which this occurs is known as the percolation point. A model of the conductivity of polymers filled with conductive particles from work by Mamunya et al [70] is shown in Figure 4.4-1. Percolation occurs in this figure between filler volume contents of $\varphi_{c1}$ and $\varphi_{c2}$. 
Figure 4.4-1: Percolation curve for polymers filled with conductive particles [70].

The conductive matrices that form with carbon black are different from the matrices that form with discrete particles or flakes because the carbon is composed of very small particles with a diameter of around 40 nm, which adhere to one another to form larger aggregates of a size dictated by the mixing process and the amount of carbon in the material. This means that the level of mixing has a significant effect on the resulting resistivity of a carbon black mixture. All mixtures in this thesis have a two minute hand mixing period before printing. After this mixing period the mixtures produced are homogeneous and free of lumps but have not yet cured noticeably.

In this work carbon loaded rubber is stencil printed and must consequently fall within a certain viscosity range. 20-100 Pa.s is used as an approximation of the required viscosity. With high loadings of carbon (>20%) the material becomes like clay, however the viscosity can be reduced with the addition of a solvent thinner. ESL T-402 and Dow Corning SL9106 are used in this work to reduce the viscosity. The effect of a reduction in viscosity on even distribution of the carbon particles can be demonstrated by stencil printing a formulation of carbon loaded rubber with the same amounts of silicone rubber and carbon black, the same mixing time and different loadings of thinner. The resistances of a set of samples composed of carbon black, T-402 thinner and RTV M silicone rubber are shown in Table 4-12. The carbon loading and thinner loadings are described in units phr (per hundred rubber). This is commonly used in the literature and states all ingredients as a percentage by weight of the amount of rubber in the mixture. The carbon loading for the test described below was 25 phr.

Table 4-12: Thinner content and resulting cured sample resistivity for RTV M silicone rubber and T402 thinner with a 25 phr carbon loading.

<table>
<thead>
<tr>
<th>Thinner (phr)</th>
<th>1</th>
<th>3</th>
<th>8</th>
<th>15</th>
<th>25</th>
</tr>
</thead>
<tbody>
<tr>
<td>Resistivity (Ω.cm)</td>
<td>1800</td>
<td>820</td>
<td>140</td>
<td>80</td>
<td>24</td>
</tr>
</tbody>
</table>

Including more thinner reduces the resistivity of the resulting mixture. Although the amount of carbon relative to the total volume in the uncured mixture is falling as more thinner is added,
the thinner is evaporated during curing so does not alter the proportions of ingredients in the vulcanised rubber. The inclusion of the thinner also reduces the viscosity and improves mixing so that the carbon is evenly dispersed throughout the mixture. Consequently, a fall in volume resistivity is effected by decreasing the viscosity.

4.4.3 Comparison between thinners for carbon loaded rubber

The inclusion of a thinner decreases resistivity and improves compatibility with stencil printing. However, it will cause shrinkage during curing as the solvent is evaporated. This can affect the durability of a textile printed with this material, because its shrinkage relative to a dimensionally stable substrate will cause mechanical forces that work against its adhesion to the substrate. Two stencil printed carbon loaded rubber dumbbell samples composed of Silastic RTV M silicone rubber and Ensaco 250G are shown in Figure 4.4-2. These two samples are from the same stencil. The sample on the left had 0 phr T-402 content and the sample on the right had 30 phr T-402 content. The weight difference of the cured samples corresponded with the amount of thinner added, indicating that all of the thinner evaporates during the curing process.

Figure 4.4-2: Difference in size between two samples with different thinner loadings. Left: 0 phr T-402 thinner. Right: 30 phr T-402 thinner.

A different thinner, Dow Corning SL9106, was trialled to attempt to reduce shrinkage. This thinner is composed of small molecules that terminate chemical bonds in the silicone that would otherwise crosslink. Consequently it does not evaporate during curing. The termination of these bonds, however, results in slower cure and reduced strength. The required carbon loading per hundred rubber is also increased when significant amounts of SL9106 are used, because it decreases the amount of carbon as a percentage of total volume or weight of the vulcanised material.

Screen printed samples of RTV M silicone rubber with a 15 phr carbon black loading and varying thinner loadings were printed on an alumina substrate using the interface. Figure 4.4-3 shows the measured resistivity of these screen printed samples, and also notes the effective carbon loading if the SL9106 content is taken into account.
Figures 4.4-3 and 4.4-4 show the resistances of screen printed samples at varying carbon loadings and a 25 phr ESL T-402 thinner loading to demonstrate the effect that the reduced carbon percolation will have on the resistance. Per square resistance is given because these samples were screen printed. The thickness was not measured but it is assumed to be similar for different formulations as the same screen was used to print each.

\[\text{Resistivity (}\Omega/\text{square})\]

<table>
<thead>
<tr>
<th>Effective Carbon Loading</th>
</tr>
</thead>
<tbody>
<tr>
<td>15</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Thinner Loading</th>
</tr>
</thead>
<tbody>
<tr>
<td>SL9106</td>
</tr>
</tbody>
</table>

This test shows again that ESL T-402 reduces the resistivity of the resulting mixture. SL9106 greatly increases the resistance of the resulting mixture. Using no thinner provided acceptable resistances in some cases but the deviation between different samples was high because these mixtures had high viscosity and were consequently less homogeneous after mixing.

ESL T-402 will therefore be used in this thesis to improve mixing. However, it must be used in relatively small amounts, so that most of the thinner evaporates before the material is printed and relatively little shrinkage results. The amount that should be used will be defined as this section progresses.

Suitable results were also achieved with a mixture of 10 phr each T-402 and SL9106 and this remains an option to be explored more fully if required. It should be noted that SL9106 also reduces adhesion which is already a common problem in the use of silicone rubbers with polyurethane [71].

### 4.4.4 Resistivity of carbon loaded rubbers

The resistivity of carbon loaded rubber was calculated by creating a set of stencil printed samples with each silicone rubber and varying carbon loadings. For each silicone rubber type, carbon loadings of 10-30 phr were used in increments of 5 phr. A T-402 thinner content of up to 10 phr was permitted if necessary to facilitate the mixing of a formulation. Each sample was cured as indicated in Table 4-11. Five samples, of thickness 1.2 mm and length and width 50 x 16 mm, were fabricated for each silicone rubber at each carbon loading. The results are shown in Figure 4.4-5. Samples that were non-conductive, insufficiently viscous to stencil print or too
viscous to mix are omitted from the results. The upper limit of resistivity as defined in the requirements is 520 Ω.cm.

It was found that Silastic RTV M provided conductivity at the lowest loading, 10 phr, while the other silicones remained non-conductive at this level of carbon loading. SilGel 612 and Viscolo 22 had similar percolation characteristics, while Sylgard 184 remained non-conductive.

One reason RTV M shows lower percolation on this chart is that it has a 36 % quartz content, which increases its specific gravity to 1.29, where the other silicones are all below 1.05. However, even if the chart were adjusted for volume rather than weight it would not explain all discrepancies. The electrical behaviour of the quartz may also have an effect. It is also possible that the intensity of the vulcanisation process has an effect on the aggregation of carbon black within the mixture. These results suggest that longer chain, flexible rubbers used in mould making, such as Silastic RTV M and Viscolo 22, provide a higher conductivity than shorter chain, more stiff silicones used in protecting electronics, such as Sylgard 184. More silicones would need to be tested before this can be confirmed but it is outside the scope of this thesis.

4.4.5 Viscosity of carbon loaded rubbers

The viscosity of a carbon loaded rubber formulations is difficult to measure on the standard cone-plate viscometer used in previous measurements. This is because the viscometer is only accurate for viscous liquids. If a high loading of carbon causes a silicone mixture to have the consistency of clay, the viscometer will give a low viscosity reading because the cone slides over the surface of the material and encounters less torque than with a viscous liquid. This means a low viscosity reading is given for a material that is observably more difficult to mix than other formulations. An example of this can be provided by viscosity measurements of Silastic RTV M with increasing carbon loadings. The measurements for nine different mixtures are shown in Figure 4.4-6.
Figure 4.4-6: Measured carbon loaded RTV-M paste viscosity on Brookfield CAP10 with different carbon loadings.

For all three formulations, the viscosity observed increased significantly with the addition of each 5 phr of carbon, but this is not shown in the viscometer readings. The Brookfield CAP10 cone-plate viscometer is therefore considered unreliable for accurately determining whether a formulation is stencil printable. Instead, from working with the materials, the amount of thinner required for sufficiently low resistivity at carbon loadings based on the results of section 4.4.4 must be determined. This was achieved by repeated attempts at formulation and stencil printing. The stencil pattern was a rectangle with width 16 mm and length 80 mm, the stencil had a thickness of 1.2 mm and the resistance was measured across the 80 mm length from the middle of each 16 mm side.

Table 4-13: Carbon loadings for stencil printable formulations of four silicone rubbers.

<table>
<thead>
<tr>
<th>Paste</th>
<th>SilGel 612</th>
<th>Sylgard 184</th>
<th>Viscolo 22</th>
<th>RTV-M</th>
</tr>
</thead>
<tbody>
<tr>
<td>Carbon loading (phr)</td>
<td>15</td>
<td>21</td>
<td>16</td>
<td>12.5</td>
</tr>
<tr>
<td>Thinner required (phr)</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>10</td>
</tr>
<tr>
<td>Resulting resistivity (Ω.cm)</td>
<td>230</td>
<td>270</td>
<td>240</td>
<td>160</td>
</tr>
</tbody>
</table>

These results are approximate, however they are useful in determining which silicone rubbers require a large amount of thinner and will have high shrinkage if used as an electrode bulk material. It was found that out of the silicones tested only RTV-M required thinner due to its high viscosity. At higher carbon loadings of around 24 phr both Sylgard 184 and Viscolo 22 require a thinner to be properly mixed. All the rubbers examined can be formulated into stencil printable pastes with resistances below the 520 Ω.cm target.

4.4.6 Tensile behaviour of carbon loaded rubbers

Tensile testing is necessary to find whether a carbon loaded rubber formulation is appropriate as a flexible electrode material. A lower Young’s modulus is preferable as this will mean the electrode is more flexible and conforms to the skin. It will also reduce the stress on the adhesive bond between the silicone and screen printed textile during flexure. A high elongation at break is useful to prevent rupture. First, to examine the effect of carbon and
thinner addition, tensile tests were performed with Viscolo 22 at varying loadings of carbon black and a 10 phr thinner loading. Standard dumbbell samples were stencil printed. Four samples were tensile tested for each of the three levels of carbon loading; 0, 10 and 20 phr. The averaged results for the three carbon loadings are given in Figure 4.4-7.

![Graph showing tensile testing of Silastic RTV M with no thinner and varying carbon loadings.](image)

Figure 4.4-7: Tensile testing of Silastic RTV M with no thinner and varying carbon loadings.

It is clear that adding carbon changes the tensile properties of this silicone rubber. The tensile tests of samples without carbon loading show a lower stiffness than the carbon loaded samples initially, but a higher stiffness at greater extensions. The dumbbell samples that had added carbon are more linear-elastic. The full results from this test are summarised in Table 4-14.

Table 4-14: Tensile properties of Silastic RTV M silicone rubber with varying carbon loading and no thinner.

<table>
<thead>
<tr>
<th>Carbon loading (phr)</th>
<th>0</th>
<th>10</th>
<th>20</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elongation at break (%)</td>
<td>791.3</td>
<td>464</td>
<td>439.9</td>
</tr>
<tr>
<td>Standard deviation (%)</td>
<td>38.7</td>
<td>14.7</td>
<td>51.3</td>
</tr>
<tr>
<td>Tensile strength (MPa)</td>
<td>1.78</td>
<td>1.10</td>
<td>1.34</td>
</tr>
<tr>
<td>Standard deviation (MPa)</td>
<td>0.15</td>
<td>0.14</td>
<td>0.19</td>
</tr>
<tr>
<td>Young’s modulus at break (MPa)</td>
<td>0.26</td>
<td>0.30</td>
<td>0.39</td>
</tr>
<tr>
<td>Young’s modulus, first 100% (MPa)</td>
<td>2.7</td>
<td>6.3</td>
<td>7.7</td>
</tr>
</tbody>
</table>

The tensile strength of cured rubbers mixed with each of the two thinners was also examined. Dumbbell samples were fabricated using RTV-M silicone rubber with a 10 phr carbon loading and 0, 10 and 30 phr loadings of T-402 and SL9106. Four samples were tested for each formulation. The averaged results from tensile tests with these five different carbon formulations are shown in Figure 4.4-8.
The material properties extracted from this test are shown in Table 4-15.

Table 4-15: Tensile properties of Silastic RTV M silicone rubber with 10 phr carbon loading and varying thinner loading.

<table>
<thead>
<tr>
<th>Thinner type</th>
<th>None</th>
<th>ESL-T402</th>
<th>Dow Corning SL-9106</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thinner loading (phr)</td>
<td>0</td>
<td>10</td>
<td>30</td>
</tr>
<tr>
<td>Elongation at break (%)</td>
<td>210.42</td>
<td>266.99</td>
<td>227.50</td>
</tr>
<tr>
<td>Standard deviation (%)</td>
<td>12.62</td>
<td>12.66</td>
<td>32.3</td>
</tr>
<tr>
<td>Tensile strength (MPa)</td>
<td>4.34</td>
<td>3.34</td>
<td>2.32</td>
</tr>
<tr>
<td>Standard deviation (MPa)</td>
<td>0.17</td>
<td>0.17</td>
<td>0.35</td>
</tr>
<tr>
<td>Young's modulus</td>
<td>2.06</td>
<td>1.25</td>
<td>1.02</td>
</tr>
</tbody>
</table>

This test shows that the tensile properties are altered with the addition of thinner whichever thinner is used. T-402 moderately decreases the tensile strength at a 30 phr loading, but has no observable effect on it at a 10 phr loading. SL9106 reduces the tensile strength more severely than T-402 at 10 and 30 phr loadings. This shows that SL9106 inhibits crosslinking more severely than T-402, but that T-402 appears to have an effect on the curing process at higher loadings.

The elongation at break is increased with T-402 contents of 10 and 30 phr, but significantly decreased with a 30 phr SL9106 loading. Again, this corroborates with more inhibition of vulcanisation with SL9106 than with T-402.

Following these tests, it was necessary to tensile test the carbon loaded rubber formulations determined by the resistivity and printability experiments. This will show how the various formulations behave under mechanical stress. Four standard dumbbell samples were stencil printed for each formulation. The carbon loaded rubber formed with SilGel 612 had very low tensile strength and consequently was excluded from the selection process and not tested.
The results of the tensile test for carbon loaded rubber formulations based on the other three silicone rubbers are shown in Figure 4.4-9.

![Graph showing tensile test results](image)

Figure 4.4-9: Extension of three carbon loaded rubbers using different silicone bases in formulations to give similar resistivity. Carbon loading is given in the legend.

The resistivities of the samples used in this tensile test are given in Table 4-16.

Table 4-16: Carbon loadings and resistivities of dumbbell samples used in the above tensile test.

<table>
<thead>
<tr>
<th>Base</th>
<th>Sylgard 184</th>
<th>Viscolo 22</th>
<th>Silastic RTV M</th>
</tr>
</thead>
<tbody>
<tr>
<td>Carbon loading (phr)</td>
<td>22</td>
<td>16</td>
<td>12.5</td>
</tr>
<tr>
<td>Resistivity (Ω.cm)</td>
<td>270</td>
<td>240</td>
<td>160</td>
</tr>
<tr>
<td>Standard deviation (Ω.cm)</td>
<td>50</td>
<td>40</td>
<td>40</td>
</tr>
<tr>
<td>Tensile strength (MPa)</td>
<td>0.79</td>
<td>1.01</td>
<td>1.93</td>
</tr>
<tr>
<td>Elongation at break (%)</td>
<td>87.3</td>
<td>398.8</td>
<td>161.3</td>
</tr>
<tr>
<td>Young’s modulus</td>
<td>0.91</td>
<td>0.25</td>
<td>1.20</td>
</tr>
</tbody>
</table>

It is clear from these results that Viscolo 22 has the most appropriate tensile characteristics for use in a flexible system. It has a lower Young’s modulus and consequently will not add excessive stresses on the printed system during flexure. It also has a significantly higher elongation at break.
4.4.7 Selected carbon loaded rubber formulation

Table 4-17 shows a summary of the carbon loaded rubber paste properties.

Table 4-17: Summary of experimentally obtained characteristics for four silicone rubbers.

<table>
<thead>
<tr>
<th>Paste</th>
<th>SilGel 612</th>
<th>Sylgard 184</th>
<th>Viscolo 22</th>
<th>RTV-M</th>
</tr>
</thead>
<tbody>
<tr>
<td>Carbon loading required for minimum conductivity (phr)</td>
<td>-</td>
<td>22</td>
<td>16</td>
<td>12.5</td>
</tr>
<tr>
<td>Thinner loading required for printability (phr)</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>Curing time (minutes)</td>
<td>15</td>
<td>45</td>
<td>10*</td>
<td>20</td>
</tr>
<tr>
<td>Curing temperature (°C)</td>
<td>100</td>
<td>100</td>
<td>100*</td>
<td>80</td>
</tr>
<tr>
<td>Mixed viscosity (Pa.s, 0 phr)</td>
<td>1</td>
<td>3.9</td>
<td>4</td>
<td>90</td>
</tr>
<tr>
<td>Tensile strength (MPa)</td>
<td>-**</td>
<td>0.79</td>
<td>1.01</td>
<td>1.93</td>
</tr>
<tr>
<td>Elongation at break (%)</td>
<td>-**</td>
<td>87.3</td>
<td>398.8</td>
<td>161.3</td>
</tr>
<tr>
<td>Young’s modulus</td>
<td>-**</td>
<td>0.91</td>
<td>0.25</td>
<td>1.20</td>
</tr>
</tbody>
</table>

* Determined experimentally, others are from datasheets.
** SilGel 612 could not be tensile tested as its tensile properties were very poor.

Viscolo 22 has been selected as the electrode bulk material. It has the lowest Young’s modulus and can be formulated into a stencil printable paste with cured resistivity significantly under the 520 Ω.cm requirement without the need for a thinner. At a later stage in the project a carbon loading of 20 phr and a T-402 loading of 10 phr was adopted as this achieved better conductivity (100-150 Ω.cm) with a similar viscosity, however no further methodical selection process was carried out. No problems with shrinkage have been noted with this level of thinner loading and it is assumed that most of it evaporates during the mixing process. This formulation is denoted in this thesis as the standard carbon loaded rubber formulation.

4.5 Conclusions

In this section screen printable pastes for the interface and encapsulation and the conductive layers were selected. A stencil printed carbon loaded rubber formulation for use in electrodes was developed.

More sophisticated techniques could be substituted for most of the experiments carried out in this section, but those performed here are appropriate to the time and resources that could be devoted to material selection and formulation within the context of this research. The experiments were sufficiently detailed and accurate to define the important properties of each material.

UoS-IF-#4 and UoS-IF-#39 both have useful properties for use as an interface and encapsulation material. UoS-IF-#4 has high tensile strength and high surface energy, but absorbs water. UoS-IF-#39 does not absorb water and has a significantly higher elongation at break, but has low surface energy and consequently printed pastes adhere poorly to its surface. The rheology of both is appropriate for screen printing. These two pastes will be examined further in Chapter 6.
The commercial silver pastes examined had good resistance to abrasion and high durability after deformation. The most durable paste examined was Electrodag 725A, followed by Conductive Compounds Ag-800 and then Dupont 5000. Electrodag 725A is the most suitable paste of those examined, however Dupont 5000 will be used in investigations into the durability of different structures as changes in resistance will be more obvious. The University of Southampton silver paste, UoS-TC-32, will not be used as it had poor durability and poor abrasion resistance.

The selected carbon loaded rubber formulation will be composed of Viscolo 22 silicone rubber, Ensaco 250G carbon black, and T-402 thinner. The carbon black and thinner will be used in amounts 20 phr and 10 phr respectively. This formulation offers the best combination of stencil-printability, conductivity and flexibility, and has no visible shrinkage after vulcanisation.
5 - Fabrication process

5.1 Introduction
In the previous two chapters designs have been given and materials have been selected for the fabrication of screen and stencil printed biopotential monitoring networks on textiles. This chapter describes the fabrication process for these networks. The fabrication of the textile monitoring system is split here into several stages, each discussed in its own section.

- Textile preparation
- Screen printing
- Electrode fabrication
- Sewing
- Via fabrication

5.2 Textile preparation
Samples of textile are cut in to the required dimensions. For most experiments described here the textile is cut into squares of 150 x 150 mm. Each textile sample is ironed before printing to remove any creases which may have formed in the textile. After ironing, the textile samples are glued to alumina tiles of the same dimensions and thickness 0.635 mm for the duration of the print and processing. This provides a rigid backing for easy handling and consistent alignment of the textile during the printing process. The glue used is 3M spray mount repositionable adhesive which allows the textile to be removed easily from the tile after processing with the application of heat. 80 °C for 10 minutes is used for all removal processes in this thesis. The textile substrates are then shaved with a Remington rotary electric shaver to remove any fibres protruding from the textile, reducing the pilosity of the textile and improving the printability of its surface. This technique was developed at the University of Southampton during the Microflex project [72].

5.3 Screen printing
In this work the textiles are screen printed with a DEK 248 flatbed printer. This has a maximum printable area of 280 x 280 mm using standard screens although larger screens can provide a printable area of up to 500 x 500 mm. Smaller screens used at the University of Southampton, which are used in most of the work described herein, have a printable area of 200 x 200 mm. All screens have a mesh density of 101 threads per centimetre and are made of stainless steel. Each thread in the mesh has a diameter of 80 µm. The thickness of the emulsion on the screen, which forms the print pattern, dictates the thickness of the deposited layer. Emulsion thicknesses used in this thesis are 5 µm for the conductive layer and 30 µm for interface and encapsulation layers. The squeegee used is made of polyurethane, has a shore hardness of 70-75 and a print angle of 45°. The substrate to be printed is on a substrate tray, or platen, that slides into the printer for printing and slides out again to allow the substrate to be removed and cured. The DEK 248 is shown in Figure 5.3-1.
The screen printing procedure can be described in several steps. These are:

- **Setting up**: Ensuring the screen is free from contamination and inserting it into the printer. The substrate is placed on the substrate tray and the tray is slid into the printer so that the substrate is beneath the screen. For the first layer of a design, the substrate tray is aligned to the screen so that the design is centred on the textile. For subsequent layers, the substrate is aligned so that the newly printed layer is positioned correctly relative to previous layers. The spacing between the screen and substrate is set by the user.

- **Pasting the screen**: The paste to be printed is placed on top of the screen by hand. The amount of paste required for a given design size and shape is determined through experience. Paste is placed on the screen at each end of the print pattern to give good coverage without the risk of paste leaking through the print pattern before printing takes place. A screen pasted with UoS-IF-#4 is shown in Figure 5.3-2.

- **Flooding the screen**: A stainless steel flood blade is used to spread the paste over the screen in a thin layer. Having a thin, even covering of paste over the screen ensures an even deposit. The gap between the flood blade and the screen is set by hand. A flooding process is shown in Figure 5.3-3.

- **Layer deposition**: The polyurethane squeegee drops on to the flooded screen with a user-determined print pressure and slides back up the screen at a user-determined speed. This pushes paste through the patterned gaps in the screen on to the substrate. A layer deposition process is shown in Figure 5.3-4.
There are several parameters that can be set differently for each layer. The print gap between screen and substrate, the squeegee pressure and the print speed are the three most important parameters in setting appropriate print conditions for each layer. The print gap, squeegee pressure and print speed used in this thesis are 0.9 mm, 6 kg and 50 mm/s respectively unless stated otherwise.

To demonstrate how layers are built up to form a printed structure on textiles, the printing process shown here was carried out with a set of screens designed to fabricate a pair of electrodes, conductive tracks and via terminations. This is denoted as screen EL-1. The layer design corresponding to each screen is shown in Figure 5.3-5.
5.3.1 Interface layer

The interface layer is formed in around four print-cure cycles (prints) of UoS-IF-#4 or UoS-IF-#39 with a screen emulsion of 30 µm. The exact number of prints varies between different textiles but the resulting interface layers in this thesis have a thickness between 100-200 µm. Each print can have one or several deposits. The number of deposits and prints required to provide the interface layer is dependent on the roughness and porosity of the textile. A very rough textile will require more prints before the surface is even, and a porous or absorbent textile will allow more paste to sink in to the textile. Some porosity is useful as the paste that has sunk in to the textile bonds mechanically to the textile when cured. The printed UoS-IF-#4 interface layer on Lagonda for design EL-1 is shown in Figure 5.3-6.

![Figure 5.3-6: A screen printed UoS-IF-#4 interface layer on Lagonda using design EL-1.](image)

This interface layer requires 3 prints composed of 6 (1<sup>st</sup> print), 2 (2<sup>nd</sup> print) and 1 (3<sup>rd</sup> print) deposits. More deposits are required on the first layer as some UoS-IF-#4 soaks into the textile initially.

UoS-IF-#4 is used for all screen printed interface and encapsulation layers described in this chapter as it is easy to use and these samples are not subjected to durability tests. The use of UoS-IF-#39 to improve durability in wet conditions is described in chapter 6.

5.3.2 Conductive layer

A layer of conductive paste is printed on to the cured interface layer. If the interface layer has been printed properly and provides a smooth surface, there are rarely issues with this step. A single deposit of silver paste is deposited on to the surface. The screen for this layer has an emulsion thickness of 5 µm resulting in a layer thickness of 5-10 µm. This was cured at 120 °C for 10 minutes. The interface layer and a single deposit of Dupont 5000 silver polymer paste are shown in Figure 5.3-7.
5.3.3 Encapsulation layer

The conductive paste is encapsulated with a second layer of polyurethane paste. Again, if previous layers have been printed correctly, there are rarely issues with this step. A single deposit is sufficient to protect the printed silver from abrasion; however more layers can be used to increase the durability during bending normal to the plane. Screens with an emulsion thickness of 30 µm are used for encapsulation layers. Two deposits of UoS-IF-#4 are printed here to encapsulate the conductive tracks, resulting in a layer approximately 35 µm thick. The completed screen printed conductive track structure is shown in Figure 5.3-8.

The encapsulation layer seals in the cured silver polymer at the conductive track. The termination pads are left exposed to form electrodes and vias. The silver polymer layer and the polyurethane interface and encapsulation layers can be distinguished when the cross section of the conductive track is observed. A narrow printed conductive track printed on Escalade was cut with scissors and the cross section is observed with a Zeiss Evo SEM in Figure 5.3-9.
5.3.4 Electrode Pads

It is possible to screen print electrodes using the standard carbon loaded rubber formulation described in the paste selection chapter. This carbon loaded rubber layer protects the silver from being exposed to the skin, reducing the likelihood of biocompatibility issues. It also protects the silver itself from abrasion.

A slightly modified formulation, consisting of Viscolo 22, 18 phr 250G carbon black and 20 phr T-402 thinner, was originally used in order to ease the screen printing process. However, this material is not ideal for screen printing because it begins to vulcanise as soon as it is mixed and will clog up the screen over a printing period longer than an hour. These screen printed electrodes were eventually discarded in favour of stencil printed electrodes to increase the thickness that could be printed in a single deposit and thereby speed up fabrication. Screen printed carbon loaded rubber electrodes are shown in Figure 5.3-10.

5.4 Passive electrode fabrication

Stencil printing is used to fabricate the thick, carbon loaded rubber structures that form the bulk of passive electrodes in this thesis. The stencils are made from 3 mm thick aluminium
The screen printed textile is aligned with the stencil and clamped between the stencil and another piece of aluminium, to ensure it does not move relative to the stencil during printing. The carbon loaded rubber paste is then deposited into the stencil pattern with a spatula. A polyurethane squeegee, as used in screen printing, is slid across the stencil to remove excess paste from the stencil surface. The stencil printing setup is shown in Figure 5.4-1.

Figure 5.4-1: The stencil printing setup for fabricating thick carbon loaded rubber structures on textiles.

The clamped stencil is then placed into the oven at 80 °C for 30 minutes, after which the silicone has vulcanised and consequently the electrodes have solidified. The clamps are then removed and the bottom aluminium sheet is removed. The electrodes can then be pushed out of the stencil to release the printed structure. The screen and stencil printed textile is then placed back into the oven with a weight depressing the electrodes to improve their adhesion. A further 10 minutes at 80 °C ensures a full cure and acceptable adhesion.

Several of the stencils used were chamfered, as it was thought this would improve durability and cause the carbon loaded rubber to bond with the textile. However, this was not an appropriate choice with the fabrication techniques used here. Although some mechanical bonding was achieved between the textile and the carbon loaded rubber, generally the chamfered stencils had poor edge definition compared to the perpendicular stencils, as the carbon loaded rubber did not fill the chamfered space evenly. Screen and stencil printed electrodes were printed on to electrode pads manufactured with screens EL-1. These are shown in Figure 5.4-2 and Figure 5.4-3 respectively. Both of the electrodes shown have skin contact area 20 x 20 mm.
5.5 Active electrode fabrication

The active electrode fabrication process involves four distinct stages. These are screen printing, component attachment, component encapsulation and electrode stencil printing. This process provides an active electrode with self-contained electronics that will be subject to less deformation compared with placing electronics external to the electrode.

5.5.1 Active electrode screen printing

The circuit connections for the active electrode are printed with the same process that was previously described, and are printed at the same time as the conductive tracks between the active electrodes and vias. This design provides pads for the placement of two resistors, a capacitor and an op-amp as used in the circuit described by Kang et al. The screen printed portion of an active electrode on Lagonda, printed using UoS-IF-#4 interface and encapsulation paste and Dupont 5000 silver polymer paste, is shown in Figure 5.5-1.

5.5.2 Active electrode component attachment

After screen printing, electronic components are directly attached to the screen printed silver polymer circuit layer. A conductive epoxy, RS silver loaded epoxy adhesive, is stencil printed with the 100 µm thick stencil shown in Figure 5.5-2. This epoxy is used as it cures at a relatively low temperature 80 ºC for 30 minutes and so does not damage the textile. The components
are then placed by hand on to the stencil printed conductive epoxy pads and left to dry. It is important that the printed structure is not flexed at this stage because the electronic components can easily detach from the bonding pads while the epoxy is drying. After the epoxy has dried the active electrode circuit is fully functional. The active electrode circuit on textile with components attached is shown in Figure 5.5-3.

5.5.3 Active electrode component encapsulation

The components are encapsulated with a stencil of thickness 3 mm. The height of the op-amp, the tallest component, plus the conductive epoxy on the bonding pads is around 1.5 mm so a stencil of height 2 mm would be preferable to improve the compactness of the electrode. 3 mm was used because it was the closest thickness available in stock aluminium that could be machined into a stencil.

Both UoS-IF-#4 and UoS-IF-#39 have been trialled for use as this encapsulation material. Circuits encapsulated with UoS-IF-#39 were prone to circuit failure. It is believed that this is due to deformations at the circuit due to the flexibility of this material, although this has not been shown experimentally. UoS-IF-#4 is the preferred choice because it is stiffer, with a significantly higher Young’s modulus as demonstrated in chapter 4. Circuits encapsulated with UoS-IF-#4 have not failed over the course of this research.

These two polyurethane materials were trialled as circuit encapsulations because they had good adhesion to screen printed polyurethane layer. They cure under exposure to UV light. The curing energy is not known as this is carried out with a UV pen (UV-P 280) rather than the UV light chamber. Other materials may offer better protection to the circuit and a full investigation into the use of these materials in this application is required to evaluate their suitability. It may be the case that more traditional circuit encapsulation materials such as the previously mentioned Sylgard 184 are more appropriate, although this material would have lower adhesion to the screen printed layer. The stencil is shown in Figure 5.5-4 and the active electrode with an encapsulated circuit is shown in Figure 5.5-5.
5.5.4 Active electrode stencil printing

The final stage in the fabrication of the active electrode is the encapsulation of the electrode in carbon loaded rubber. This creates an electrical path from the skin surface to the input of the active electrode circuit. This is cured at 80 °C for 30 minutes using the same process as described for the fabrication of passive electrodes. The aim was to use a chamfered stencil to ensure a mechanical bond between the textile around the base of the electrode and the carbon loaded rubber used for the electrode bulk. Although this bond was created in places, it was weak as there was minimal penetration of the carbon loaded rubber into the textile. The manual stencil printing process also meant that the edge definition was poor. The surface of the electrode is rough however this is not necessarily a problem in biopotential electrodes. It has been suggested that a rough surface tends to improve the performance because, since there are hairs on the skin, more of the surface makes contact with the skin than would with a smooth electrode surface. The chamfered stencil is shown in Figure 5.5-6 and the finished electrode is shown in Figure 5.5-7.
5.6 Sewing

Sewing is necessary for the correct operation of the devices described in this thesis. It allows the printed inelastic textiles created using the fabrication processes described up to this point to be integrated into functional garments. By sewing the printed inelastic textiles on to a strip of more elastic textile, a band is created that can be wrapped around the part of the body that is to be monitored. Sewing is also used to attach hook and loop fastening textile to the ends of the strip so that the functional garment can be secured in place once in the correct position. Sewing together the printed inelastic, the elastic and the hook and loop textiles results in a wearable set of electrodes. A set of electrodes on a chest band for one-lead bipolar ECG monitoring is shown in Figure 5.6-1.

![Image of a wearable chest band](image)

Figure 5.6-1: The printed inelastic, unprinted elastic and hook and loop textiles sewn together to form a wearable chest band for one-lead bipolar ECG.

Although other methods to attach textiles were investigated, such as ‘iron-on’ hook and loop textile and bonding with silicone rubber, it was found that sewing was the quickest, cheapest and most durable method for attachment between different textiles. The sewing machine used in this work is a Brother X-5. Although hook and loop textile is useful for prototypes it tends to gather other textile and is not durable in the long term. Other methods of securing the printed textile around the body should be examined in future.

5.7 Via fabrication

The via fabrication is completed in two stages. First, the via is created by inserting a steel button through the screen printed termination. Second, the button is encapsulated on the inside of the garment to avoid any unwanted contact between the skin and the conductive path. An example where six vias are created at the terminations of three conductive tracks is provided. The three conductive tracks are printed on Lagonda and sewn on to elastic fabric, as shown in Figure 5.7-1.
5.7.1 Via creation

The vias are created by clamping two-part steel buttons through the screen printed track terminations. By removing one of the five prongs on the steel button, it is ensured that there is no puncturing of the conductive track where it meets the termination pad. The remaining four prongs hold the two parts of the button together tightly, clamping the upper part to the screen printed silver layer and ensuring a reliable electrical connection through the textile. This allows a connection to an amplifier or other external electronics to be made from the reverse side of the textile and prevents wires being placed next to the skin. The front and back of the textile after the via creation stage are shown in Figure 5.7-2 and Figure 5.7-3 respectively.

5.7.2 Via encapsulation

The front side of the via connections are encapsulated to avoid unwanted electrical connection with the skin at this point. This also prevents the silver polymer layer from being exposed to abrasion or moisture. The back side is left unencapsulated so that the button can be connected to wires.
UoS-IF-#39 paste is used for these encapsulations because, unlike the active electrode circuit encapsulation discussed on page 93, they are exposed to the skin. This material is softer than UoS-IF-#4 while retaining the adhesion advantages of a compatible polyurethane paste. This cures under exposure to UV light. The exact curing energy used is not known as this is carried out with a UV pen (UV-P 280) rather than the UV light chamber.

Two encapsulation procedures are trialled in this thesis, stencil printed and glob-top. The stencil printing procedure is similar to that described for passive electrode fabrication in terms of the stencil and clamp rig used. This involves 30 seconds of UV curing, removal of the stencil, then a further 30 seconds of UV curing. Glob top encapsulation, by contrast, does not require a stencil and the paste is deposited on to the surface with a spatula or syringe and cured for 60 seconds to solidify it and prevent it spreading. Completed sets of stencil printed and glob top encapsulated vias are shown in Figure 5.7-4 and Figure 5.7-5 respectively.

![Figure 5.7-4: Stencil printed encapsulation for washing test samples.](image)

![Figure 5.7-5: Glob-top encapsulation for washing test samples.](image)

5.8 Conclusions

A fabrication process has been described that can be used to produce networks of passive or active electrodes on woven textiles. This process is homogeneous, involving printing for every stage apart from the component and via placement. In this process stencil printing is used for component encapsulation and carbon loaded rubber electrode deposition, however this could be replaced with dispenser printing, 3D printing or injection moulding. This would improve the suitability of the technology for continuous production and also solve the problem of poor edge definition on the stencil printed electrodes. Further, it could also improve the mechanical bonding between the textile and the carbon loaded rubber as discussed in appendix D.

The processes described in this chapter can be used to fabricate all the component designs described in chapter 3 using the materials selected or formulated in chapter 4. In the remainder of this thesis fabrication processes are described only where they deviate from those described in this chapter.
6 Conductive track durability

6.1 Introduction

This chapter presents an investigation into design issues regarding the layer structure of the screen printed conductive paths. The objective of this chapter is to maximise protection of the cured silver polymer layer within the printed structure by selection and orientation of textile, waterproofing, and optimising the layer structure.

First, the required thickness of an interface layer on three different textiles is examined. Conductive track structures composed of UoS-IF-#4 printed on textile are mandrel tested and machine washed to gauge their durability and their design is refined. Then, structures composed of layers of UoS-IF-#4 and UoS-IF-#39 are mandrel tested to find the most durable. Once the approximate layer structure has been determined a further mandrel test on variations of this structure is carried out to find the optimal structure that can be printed. These are machine washed to gauge their durability.

Dupont 5000 silver paste is used to fabricate the conductive layer for these experiments because it shows the worst durability out of the commercial silver pastes previously examined and will consequently allow the differences in durability between different structures to become apparent more quickly. Mandrel tests are used to gauge the effect of repeated mechanical stress on the resistance of printed conductive tracks, isolating this from moisture and heat encountered in a washing machine. Mandrel tests show the effect of bending normal to the plane on a printed conductive track. The effect of this repeated bending is to displace the silver particles and create an uneven distribution within the cured silver paste, thereby increasing resistance. It is assumed in this section that there is a direct relationship between the strain at the silver layer and the resistance increase after repeated cycles of this strain. A model was developed as the experiments proceeded to explain the observed behaviour.

The improvements to durability were gauged with tests in a washing machine, as this is widely considered to be the highest level of mechanical stress and moisture that wearable electronics will endure. This shows which structures are waterproof and the way in which the effects of random stresses encountered in a washing machine differ from those of controlled stresses in a repeated mandrel test.

It has already been discussed that the ideal placement of the conductive track within the printed structure is along the neutral axis. However, the textile has unusual mechanical properties, in that it is able to compress with minimal force but requires a large force to extend. Consequently, the behaviour of the printed structure and the location of the neutral axis are different during bending in internal and external orientations. Internal and external bending orientations are shown in Figure 6.1-1.
These two bending orientations will be denoted with reference to the location of the printed structure. Bending with the printed structure on the inside will be referred to as internal bending and bending with the printed structure on the outside will be referred to as external bending. A mandrel radius of 3 mm is used for all experiments in this thesis. The dimensions of a printed track are denoted as shown in Figure 6.1-2.

Figure 6.1-2: A cross-sectional diagram of a printed conductive track on textile with dimensions labelled.

The amount of planar compression or extension at different points in the printed conductive track structure during a bend around a given radius depends on the bending orientation and the position of the neutral axis in that orientation. The position of the neutral axis is itself dependent on the mechanical properties of the textile and printed materials. In this chapter it is assumed that the textile is mostly extended during internal bending and mostly compressed during external bending.

6.2 Printed conductive tracks using UoS-IF-#4

In this section, printed conductive tracks with UoS-IF-#4 as the interface and encapsulation material are examined. When cured, this material had useful properties in that it had the highest surface energy, tensile strength and Young’s modulus of the interface materials examined. However, it also had the highest level of water absorption.

6.2.1 Effect of textile orientation

This experiment examined the effect of the planar orientation of a conductive track relative to the textile, specifically looking at the effect on its behaviour during bending. Printed samples were fabricated using UoS-IF-#4 as the interface and encapsulation material, Dupont 5000
silver paste as the conductor and Escalade as the textile. Screen design PC-2 was used. The interface thickness was approximately 135 \( \mu \text{m} \), the silver layer thickness was approximately 5 \( \mu \text{m} \) and the encapsulation thickness was approximately 60 \( \mu \text{m} \). The conductive tracks had length 40 mm and a conductive layer width of 1 or 2 mm with interface and encapsulation width of 1.5 or 3 mm.

Sets of samples were printed with the direction of conductive tracks aligned with either the warp or weft yarns of the textile. For each orientation, four samples were mandrel tested with the standard mandrel apparatus. Of these four, two were samples with track width 1 mm and two were samples with track width 2 mm. Testing was carried out for 200 cycles of internal bending. Each conductive track was removed from the mandrel machine periodically for resistance measurements. The results are shown in Figure 6.2-1.

![Mandrel test of encapsulated printed conductive tracks on Escalade textile, printed to align with the warp and weft orientations and subjected to 200 cycles of internal bending.](image)

The results show that aligning conductive tracks with the warp yarns causes significantly more damage to the conductive path during this test. This corresponds to a higher Young’s modulus of the textile in the warp direction, as measured on page 45. As it is more difficult to extend the textile in this direction, the printed structure must compress further to allow the structure to bend normal to the plane. This increases the dimensional change at the conductive layer and the difference is observed in the periodic resistance measurements shown in the mandrel test results. Conductive tracks in this chapter will therefore be printed in the weft direction.

### 6.2.2 Interface layer thickness

The processes required to fabricate an interface layer surface with a visibly smooth surface by printing UoS-IF-#4 on three different textiles are described in Table 6-1.
The thicker textiles have thicker yarns resulting in a rougher surface. This means more deposits are required to create a smooth interface surface on the textile. The thicker yarns also tend to give higher porosity, causing the interface to sink into the textile and meaning more interface paste is required. This corroborates previously described work by Karaguzel et al [31].

In this chapter Lagonda and Escalade are used as substrates for printed conductive tracks. The interface layer required on Elasta is thicker, and the high stretchability of this textile, examined in section 3.3, makes it unsuitable as a substrate as the printed structures can be damaged during its extension.

### 6.2.3 Encapsulation layer thickness

Although the interface layer thickness can be altered, it was decided that the methodology should be to minimise the interface layer thickness and change the encapsulation layer thickness to optimise durability. This helps to keep thickness to a minimum. To illustrate the effect on durability of changing the encapsulation layer thickness, a mandrel test was performed with printed conductive tracks in the standard mandrel apparatus.

Printed samples were fabricated using UoS-IF-#4 as the interface and encapsulation material, Dupont 5000 silver paste as the conductor and Escalade as the textile. Screen design PC-2 was used. Escalade has a thickness ($t_{tex}$) of 410 µm. The interface thickness ($h_i$) was 135 µm, the silver thickness was 5 µm and the encapsulation thickness was varied. The encapsulation thicknesses ($h_e$) were 30, 65 and 100 µm for 1, 2 and 3 prints of the encapsulation layer. The conductive tracks had lengths of 40 and 50 mm and a conductive track thickness of 1 mm. The printed structure had a width ($w_p$) of 2 mm.

Four samples were tested for each encapsulation thickness in internal and external bending orientations for 100 cycles. The printed textile conductive tracks were removed from the machine periodically to take resistance measurements. The effect of internal bending is shown in Figure 6.2-2 and the effect of external bending is shown in Figure 6.2-3.
Figure 6.2-2: Mandrel test showing the effect of internal bending on printed conductive tracks with encapsulation thicknesses of 30, 65 and 100 µm (1, 2 and 3 prints).

Figure 6.2-3: Mandrel test showing the effect of external bending on printed conductive tracks with encapsulation thicknesses of 30, 65 and 100 µm (1, 2 and 3 prints).

The resistance measurements are plotted on these graphs only while all printed tracks tested remain conductive, for example one of the 2 layer encapsulation structure samples in Figure 6.2-2 failed after 80 cycles.

In both bending orientations increasing the number of encapsulation layers improved the durability of the conductive track. A thicker encapsulation reduced the resistance change by reducing the deformation at the silver layer. It also reduced the likelihood of the track rupturing, possibly because it takes a crack longer to propagate through the thicker encapsulation. A track rupture is a crack through the printed track structure, which breaks the conductive path between its terminations, and can easily be observed with the naked eye. A micrograph of a track rupture is shown later in Figure 6.2-14.

The other conclusion from these two graphs is that there is a significant difference between the effects of internal and external bending. Internal bending causes a greater resistance increase than external bending in all of the examined structures. However, tracks that are bent externally are significantly more likely to rupture, with four ruptures out of twelve tested.
tracks in the extension tests compared to just one in the compression tests. This conclusion is corroborated by larger data sets later in this chapter.

As the textile has high tensile strength but negligible compressive strength, the strain on the conductive layer will be different in different bending orientations. In internal bending the textile is extended and the strain on the printed structure, including the silver, is greater resulting in a larger resistance increase in this bending orientation.

The greater number of ruptures in external bending is thought to be caused by the fact that the textile mesh reinforces the embedded printed layers and prevents rupture at this side. Consequently the polyurethane is more likely to rupture when the non-embedded top surface is extended, as it is during external bending.

6.2.4 Effect of silver position

In order to understand the behaviour of these structures during bending more experimental data was required. Two UoS-IF-#4 structures are examined here, with the silver layer printed at different heights throughout these structures. Structure 1 is composed of eleven prints and structure 2 is composed of seven prints. There is only one silver layer in each structure, composed of a one deposit print of Dupont 5000 with a thickness of 5 µm. The others are prints of UoS-IF-#4 layers with thickness 35 µm.

The textile used for this experiment was Lagonda because it has a lower thickness than Escalade, 0.29 mm compared with 0.41 mm, and a lower surface roughness. The lower thickness will reduce the total thickness of the structure, reducing the strain necessary for a given bending radius. Also, the reduced surface roughness means that the silver can be printed after just three layers of UoS-IF-#4, yielding seven possible configurations of the eleven print structure (structure 1) and three possible configurations of the seven print structure (structure 2). These structures are described in Table 6-2.

Table 6-2: Layout for ten printed structures, with seven varieties of structure 1 (printed thickness 350 µm) and three varieties of structure 2 (printed thickness 210 µm). All thicknesses are given in µm.

<table>
<thead>
<tr>
<th>Printing Order</th>
<th>Layer:</th>
<th>Material:</th>
<th>Layer thickness in each structure:</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>1.a</td>
</tr>
<tr>
<td>1</td>
<td>Interface</td>
<td>#4</td>
<td>100</td>
</tr>
<tr>
<td>2</td>
<td>Conductor</td>
<td>Silver</td>
<td>5</td>
</tr>
<tr>
<td>3</td>
<td>Encapsulation</td>
<td>#4</td>
<td>245</td>
</tr>
<tr>
<td></td>
<td>Printed Thickness (t_p):</td>
<td></td>
<td>350</td>
</tr>
<tr>
<td></td>
<td>Total Thickness (t_t):</td>
<td></td>
<td>640</td>
</tr>
<tr>
<td></td>
<td>Silver position (h_s):</td>
<td></td>
<td>100</td>
</tr>
</tbody>
</table>

These structures were subjected to a mandrel test of 100 cycles at a radius of 3 mm. Three conductive tracks were tested for each structure in the internal and external mandrel orientations. For external bending, ruptures in the conductive track occurred beyond 50 cycles so only the average after 50 cycles is shown here. The results are shown in Figure 6.2-4.
These results show again that internal bending, where the printed structure is mostly compressed, has a greater detrimental effect on conductivity and a greater dependence on the silver position in the printed structure than external bending. The difference arises because of the difference between tensile and compressive strength in the textile. As already mentioned, this causes greater deformation of the silver layer during internal bending which accounts for the difference in resistance change.

The printed structure cross section changes significantly along the length of the conductive track due to the rough surface of the textile. During external bending, the textile can effectively be ignored due to its low compressive strength and consequently the position of the neutral axis, where there is no deformation during flexing, varies along the length of the track and does not run parallel to the substrate. This is a possible explanation for the lack of correlation between resistance change in external bending and the height of the conductive layer in the structure. The development of this theory is continued later in this section after further experimentation provides more data.

### 6.2.5 Model of bending of UoS-IF#4 structures

A model of these structures is developed here to validate explanations for the behaviour observed in mandrel tests. It is assumed that the increase in resistance in the conductive track is proportional to the strain it experiences.

The approach used here is to assume that bending forces are always sufficient to bend the structure around a given radius, and consequently no external forces are modelled. It is assumed that because the silver layer is very thin it can be neglected in the mechanical model. It is assumed that the textile and polyurethane materials obey Hooke’s law, and that the polyurethane materials have the same Young’s modulus in extension and compression. Since there is zero strain at the neutral axis, $\bar{z}$, the strain at position $z$, $\varepsilon_z$, when the structure is bent around radius $r$, is given by Equation 1. This equation takes advantage of the fact that the neutral axis length is unchanged during bending. A diagram of the bending is shown in Figure 6.2-5.
\[ \varepsilon_z = 100 \times \left( \frac{(\text{Change in Length})}{(\text{Original Length})} \right) \]

\[ \varepsilon_z = 100 \times \left( \frac{(\text{Original Length} - \text{New Length})}{(\text{Original Length})} \right) \]

\[ \varepsilon_z = 100 \times \left( \frac{\theta.(\bar{z} + r) - \theta.(z + r)}{\theta.(\bar{z} + r)} \right) \]

\[ \varepsilon_z = 100 \times \left( \frac{\bar{z} - z}{\bar{z} + r} \right) \]

Equation 1: Calculation of strain as a percentage at point \( z \) for a structure with neutral axis at \( \bar{z} \).

Figure 6.2-5: Diagram of a printed textile structure (yellow) bent around a radius \( r \) (grey) for angle \( \theta \).

The planar strains in structures of the thicknesses used in this work are unlikely to exceed 20% at a bending radius of 3 mm. For most cases the maximum extension during bending will be around 10%. For example, if the neutral axis, \( \bar{z} \), is known to be in the centre of a 600 µm thick uniform material beam structure, at height 300 µm, the maximum strains will be at 600 µm and 0 µm. Position \( z \) is selected to be 0 µm for this example. If the bending radius, \( r \), is 3 mm, then the strain at \( z \), \( \varepsilon_z \), can be calculated using Equation 1 as described below.

\[ \varepsilon_z = 100 \times \left( \frac{\bar{z} - z}{\bar{z} + r} \right) \]

\[ \varepsilon_z = 100 \times \left( \frac{0.0003 - 0}{0.0003 + 0.003} \right) \]

\[ \varepsilon_z = 100 \times \left( \frac{0.0003}{0.0033} \right) \]

\[ \varepsilon_z = 11\% \]

In this thesis a positive value denotes compression and a negative value denotes extension. Assuming the neutral axis is close to the centre of the printed textile structure, most strains will be under 10%. This allows approximations to be taken for the tensile strength of the structural materials in the model. Young’s modulus approximations are set to fit extensions up to 10%. The straight line functions used to approximate the Young’s modulus of Escalade and Lagonda are shown in Figure 6.2-6. The full tensile test results for these textiles are shown on page 45.
Figure 6.2-6: Approximation of the tensile behaviour of Escalade and Lagonda at low extension with Young's moduli 17 MPa and 23 MPa respectively.

Figure 6.2-7 shows the Young's modulus approximation for UoS-IF-#4 and UoS-IF-#39 interface pastes. The full tensile test results for these pastes are shown on page 65.

Figure 6.2-7: Approximation of the tensile behaviours of UoS-IF-#4 and UoS-IF-#39 at low extension with Young's moduli 28 MPa and 1.5 MPa.

A model of the bending of textile printed conductive track structures was created in MATLAB using the equation for bending of a multilayer beam derived by Ballas et al [73]. It is assumed that the printed track has a rectangular cross section. For a series of layers with known widths, heights and Young’s moduli the neutral axis is given by:

\[
\bar{z} = - \frac{\sum_{i=1}^{n} \frac{w_i}{s_{11,i}} h_i^2 - 2 \sum_{i=1}^{n} \frac{w_i}{s_{11,i}} h_i \sum_{j=1}^{i} h_j}{2 \sum_{i=1}^{n} \frac{w_i}{s_{11,i}} h_i}
\]

Equation 2: Location of the neutral axis in a composite beam derived by Ballas et al [73].

Where \( n \) is the number of layers, \( w \) is the width of layer \( i \), \( h \) is the height of layer \( i \), \( s_{11} \) is the Young’s modulus of layer \( i \), and \( \bar{z} \) is the position of the neutral axis. Using this equation to find the neutral axis and the curvature with a known mandrel radius of 3 mm, the strain throughout a given structure can be predicted.
Equation 1 and Equation 2 can be used to predict the strain on the silver track by setting $z$ in Equation 1 equal to the position of the silver track in the beam and setting $\bar{z}$ equal to the value obtained with Equation 2. Inputting a set of layer parameters into the MATLAB model provides a calculation of the strain at the silver as well as the extension and compression at the extremes. A typical input parameter set is given in Table 6-3.

Table 6-3: Typical parameters for a 2-layer structure.

<table>
<thead>
<tr>
<th>Layer</th>
<th>Lagonda</th>
<th>UoS-IF-#4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Layer Number (n)</td>
<td>1</td>
<td>2</td>
</tr>
<tr>
<td>Young’s Modulus (E)</td>
<td>23 MPa</td>
<td>28 MPa</td>
</tr>
<tr>
<td>Width (w)</td>
<td>5 mm</td>
<td>3 mm</td>
</tr>
<tr>
<td>Height (h)</td>
<td>0.29 mm</td>
<td>0.35 mm</td>
</tr>
<tr>
<td>Position of Silver (z)</td>
<td>0.6 mm from base</td>
<td></td>
</tr>
</tbody>
</table>

The results so far suggest that two separate models should be used for internal and external bending. In external bending the textile is assumed to be compressed and given a Young’s modulus of zero. One issue with modelling in this fashion is that when the neutral axis position is lower than the height of the textile, different parts of the same textile will be extended and compressed and the model will use the same Young’s modulus (0 in external bending, E in internal bending) for both parts. There are clearly several possible sources of error in this model. Some assumptions have already been discussed; the silver is neglected and straight line approximations are used for the Young’s moduli of materials. The following could also cause inaccuracies in the model.

- The textile is permeable and there will be some cured polyurethane within the textile structure that should be modelled accordingly.
- The area of textile that is modelled here is not the only part of the textile that is extended, as the textile is part of a larger interconnected structure. A printed structure width : textile width ratio of 3:5 is used in the modelling described in this thesis but this is only an estimate based on visual observation.
- 0.29 mm is used as the textile thickness for Lagonda but the textile is not a rectangle with height 0.29 mm. The textile has a high surface roughness and the gaps in the surface are filled with polyurethane.

The model is developed here to improve understanding of these structures and aid design decisions. It is not meant to be an accurate method to predict the exact resistance change on a given structure during a given stress, as this is outside the scope of this thesis and such measurements can be obtained experimentally.

6.2.6 Comparison between mandrel test results and model

Modelling internal and external bending in MATLAB based on Equation 1 and Equation 2 provides an estimation of the strain during bending at a 3 mm radius for structures 1 and 2. The Young’s moduli used are 23 MPa for Lagonda and 28 MPa for UoS-IF-#4. The silver paste is neglected due to its very low thickness. The models for internal bending of structures 1 and 2 are shown in Figure 6.2-8 and Figure 6.2-9 respectively. The silver is placed in each case at a
position 100 µm above the surface of the textile, making these models equivalent to the tested structures 1.a and 2.a in Section 6.2.1.

Although the silver is at the same height above the textile in both of these structures, the strain is far greater on the silver in structure 2.a because the polyurethane layer is thinner, and consequently the neutral axis is lower in the structure and further from the silver layer. By performing the same calculations for all ten structures tested in section 6.2.4, the predicted strain at the silver from the MATLAB model can be plotted against the measured resistance change after 100 mandrel cycles in section 6.2.4. The results are shown in Figure 6.2-10.

This shows an approximately directly proportional relationship between the predicted strain from the model and the resistance change after mandrel testing. It also shows high correlation between this strain-resistance relationship in structures 1 and 2. This suggests that the internal bending model is reasonably accurate for this mandrel radius.
Modelling of external bending was also performed and the deformation at the silver was greatly reduced in modelling where the textile was given a Young’s modulus of zero, as expected. However, due to the lack of correlation between resistance change and structure in previously described mandrel tests the model did not fit the results obtained in section 6.2.4. More complex modelling is outside the scope of this thesis.

6.2.7 Washing test with UoS-IF-#4 structures

Washing machine tests are necessary to include the effects of heat and moisture in the durability testing process. Printed conductive tracks were washed until they were no longer conductive so that the mode of failure could be observed. Conductive tracks were printed on to Escalade using screen design PC-2, and sewn on to an Elasta strip backing, of width 50 mm, as printed textiles are in previously described designs for monitoring garments. Two different track widths were tested, with polyurethane widths of 1.5 and 3 mm and corresponding silver track widths of 1 and 2 mm. The interface layer and conductive layer thicknesses were kept to a minimum thickness of around 140 and 5 µm respectively and the encapsulation thickness was varied from 1 to 5 deposits, giving a thickness ranging from 35 to 170 µm. The total thickness of the printed structure above the textile ranged from 180 to 300 µm depending on the encapsulation thickness.

Five encapsulation thicknesses and two track widths gave ten different configurations for the conductive track cross section. For each configuration, nine 40 mm tracks were tested using machine washing. The nine samples of each type were cut into patches of three tracks to give 30 three-track patches, which were sewn on to six strips of elastic textile. These then had two part steel buttons clamped into the track termination pads to create vias. One of these strips is shown in Figure 6.2-11.

![Figure 6.2-11: One of six strips of Elasta with 15 printed conductive tracks.](image)

These samples were washed in a 53 minute, 40 °C cycle with spin speed 1000 rpm. The samples were dried in a box oven at 40 °C for 30 minutes after washing. The resistance of each conductive track was then measured from the vias at the back of the textile. Any track that measured above 100 Ω at any point was considered broken and no longer measured. Graphs showing the number of track breaks through ten washes for each encapsulation thickness and each width are shown in Figure 6.2-12 and Figure 6.2-13 respectively.
When a track had resistance measuring above 100 Ω there was a visible polyurethane brittle failure observable in the conductive track. After 10 washes most samples that initially measured non-conductive with a single visible break had several visible breaks. An SEM micrograph of a rupture in a printed conductive structure is shown in Figure 6.2-14.

The results show only five breaks on the first wash, but far more in the subsequent few washes. This suggests an initial weakening of some tracks, either due to their first exposure to moisture or cyclic mechanical stress. The degradation of mechanical properties after submersion, shown on page 67, would suggest that exposing UoS-IF-#4 to moisture significantly decreases the elongation at break and makes a conductive track composed of this material more likely to rupture during a washing cycle.

Increasing the track width or encapsulation thickness has benefits for the durability of the track, despite the water absorption of polyurethane. However, even the thickest track structure, with a cross section measuring 0.350 x 3 mm, had a brittle failure in four out of nine tracks tested after 10 washes. This suggests that a less brittle and more hydrophobic encapsulation material is required.

### 6.2.8 Conclusions

Mandrel testing of conductive track structures has provided various design rules for this fabrication process. First, it is clear that for printed UoS-IF-#4 structures up to a thickness of
350 µm the best position for the silver layer is as close as possible to the textile. Second, cyclic stress causes a rupture in these printed conductive tracks at extensions well below the elongation at break as measured in its tensile test.

The behaviour of conductive tracks varies dramatically between internal and external bending. The resistance change is significantly higher in internal bending. This is because the textile has greater tensile strength than compressive strength and there is consequently more deformation at the silver layer when bending involves extending the textile, in internal bending, than when it involves compressing the textile, in external bending. However, conductive tracks are more likely to rupture in external bending, as the polyurethane that is extended in this bending orientation is not reinforced by being embedded in a textile mesh.

The MATLAB model shows good correlation with the mandrel test results for internal bending. The width of the textile in the model (\(w_{\text{tex}}\)) was adjusted to account for its interconnected nature. In this model a ratio of track width to modelled textile width of 3:5 was used, however this is just an estimate and will be different for different textiles.

The Young’s moduli used in the model were approximations for this particular radius and would need to be revised for comparison with mandrel tests at a lower radius. As the bending radius decreases, the strain will increase and the linear approximation of the Young’s modulus of UoS-IF-#4 will decrease as shown in its tensile test. This will move the neutral axis further in to the textile. In internal bending at a smaller radius the effective Young’s modulus of the textile will also increase, as shown in its tensile test, having the same effect.

The washing test showed that UoS-IF-#4 was not a suitable material for encapsulating conductive tracks that are exposed to moisture because it has comparatively high water absorption. When moisture is absorbed the mechanical properties are degraded, as previously demonstrated in section 4.2.5 on page 66, and there is a dimensional change. These printed conductive tracks break too quickly during a series of machine washes and only 10% of tested tracks endured ten machine washes without failing.

It should be noted that the machine mandrel method puts strain on the whole conductive track for each cycle. In machine washing, by comparison, track bends are likely to occur less frequently and be isolated at one particular point along the track. The bending radius can be significantly lower than 3 mm.

**6.3 Printed conductive tracks using UoS-IF-#4 and UoS-IF-#39**

The previous section demonstrated that UoS-IF-#4 could not be used as the interface and encapsulation in structures that are machine washed due to the high water absorption of this material and the detrimental effect this absorption has on its tensile strength, demonstrated on page 66 of this thesis. Consequently, UoS-IF-#39 was used as an outer interface and encapsulation in order to prevent water absorption by the inner UoS-IF-#4 layers. This methodology was originally described by Yang et al [66]. The following section examines interface and encapsulation structures composed of layers of each of these materials.
6.3.1 Layer structure: UoS-IF-#4 and UoS-IF-#39

The polyurethane interface materials selected in the previous section were UoS-IF-#4, which has high water absorption, a high Young’s modulus and high surface energy, and UoS-IF-#39, which has no water absorption, a much lower Young’s modulus and lower surface energy. A mandrel test was performed on four possible designs for the interface and encapsulation layer. A diagram showing each of the designs is shown in Figure 6.3-1. These are denoted by the relative amounts of UoS-IF-#4/UoS-IF-#39.

![Figure 6.3-1: Four Interface and encapsulation layer structures. For the interface layer: (a) UoS-IF-#4 only, (b) 3 prints UoS-IF-#4, 1 print UoS-IF-#39, (c) 2 prints UoS-IF-#39, 2 prints UoS-IF-#4, (d) 1 print UoS-IF-#4, 3 prints UoS-IF-#39 only. The encapsulation layer has the same structure mirrored through the conductive layer.](image)

Conductive tracks using each layer structure were printed on to Escalade. Their thicknesses as measured with a Mitutoyo IP65 micrometer. The results are shown in Table 6-4.

<table>
<thead>
<tr>
<th>Structure</th>
<th>Total Thickness (mm)</th>
<th>S.D. (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>100/0 (0% #39)</td>
<td>0.7057</td>
<td>0.0053</td>
</tr>
<tr>
<td>75/25 (25% #39)</td>
<td>0.6795</td>
<td>0.0125</td>
</tr>
<tr>
<td>50/50 (50% #39)</td>
<td>0.6313</td>
<td>0.0072</td>
</tr>
<tr>
<td>25/75 (75% #39)</td>
<td>0.6030</td>
<td>0.0125</td>
</tr>
</tbody>
</table>

The measured thicknesses vary significantly between the different structures. It is thought that the difference arises primarily because of the low Young’s modulus of UoS-IF-#39, causing tracks with a higher proportion of this material in their structure to compress more when measured with a micrometer. It was not possible to corroborate the variation in thicknesses with SEM micrographs. These micrographs appeared to show that the structures all had the same thickness but accurate measurements were difficult to obtain as the cross sections were deformed due to the low Young’s modulus of UoS-IF-#39. The layer dimensions of these structures are described in Table 6-5.
Table 6-5: The layout of four printed structures with various proportions of UoS-IF-#4 and UoS-IF-#39. All thicknesses are given in µm.

<table>
<thead>
<tr>
<th>Printing Order</th>
<th>Layer</th>
<th>Material</th>
<th>100/0</th>
<th>75/25</th>
<th>50/50</th>
<th>25/75</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Interface</td>
<td>#39</td>
<td>0</td>
<td>35</td>
<td>70</td>
<td>105</td>
</tr>
<tr>
<td>2</td>
<td></td>
<td>#4</td>
<td>140</td>
<td>105</td>
<td>70</td>
<td>35</td>
</tr>
<tr>
<td>3</td>
<td>Conductor</td>
<td>Silver</td>
<td>5</td>
<td>5</td>
<td>5</td>
<td>5</td>
</tr>
<tr>
<td>4</td>
<td>Encapsulation</td>
<td>#4</td>
<td>140</td>
<td>105</td>
<td>70</td>
<td>35</td>
</tr>
<tr>
<td>5</td>
<td></td>
<td>#39</td>
<td>0</td>
<td>35</td>
<td>70</td>
<td>105</td>
</tr>
<tr>
<td></td>
<td>Total Thickness (t_t):</td>
<td></td>
<td>695</td>
<td>695</td>
<td>695</td>
<td>695</td>
</tr>
<tr>
<td></td>
<td>Silver position (h_s):</td>
<td></td>
<td>140</td>
<td>140</td>
<td>140</td>
<td>140</td>
</tr>
</tbody>
</table>

These samples were subjected to 200 cycles at mandrel radius 3 mm. Three conductive tracks of length 40 mm, printed structure widths 3 and 1.5 mm and conductive layer widths 2 and 1 mm were mandrel tested in internal bending for each described structure. The results are shown in Figure 6.3-2.

Figure 6.3-2: Resistance change during 200 internal bending mandrel cycles at a 3mm radius for various printed structures composed of UoS-IF-#4 and UoS-IF-#39.

These results suggest that the 50/50 structure has the best durability to internal bending at a 3 mm radius. As well as providing a waterproof outer coating, the outer layer of UoS-IF-#39 also prevents mechanical damage to the conductive layer compared to an interface and encapsulation structure of the same dimensions composed solely of UoS-IF-#4. This structure is composed of two layers of UoS-IF-#39, two layers of UoS-IF-#4 and a single deposit of Dupont 5000 silver paste. The encapsulation structure is the same as the interface structure, mirrored through the conductive layer. Even taking into account the possible height variation of the samples, these results do not fit with the model established in the previous section for predicting strain at the silver in internal or external bending. A comparison between the predicted strain at a 3 mm radius and the measured resistance change after 200 cycles at 3 mm is shown in Table 6-6.
Table 6-6: Predicted strain in the MATLAB model compared with the measured resistance change after 200 mandrel cycles.

<table>
<thead>
<tr>
<th>Structure (#4/#39)</th>
<th>100/0</th>
<th>75/25</th>
<th>50/50</th>
<th>25/75</th>
</tr>
</thead>
<tbody>
<tr>
<td>Predicted Strain (%)</td>
<td>6.86</td>
<td>7.50</td>
<td>8.26</td>
<td>9.19</td>
</tr>
<tr>
<td>Resistance Change (%)</td>
<td>748</td>
<td>534</td>
<td>428</td>
<td>558</td>
</tr>
</tbody>
</table>

It is clear that the stiffness of the beam, which decreases as the portion of UoS-IF-#39 increases, does not correlate with the resistance increase after 200 mandrel cycles. It is thought that this lack of correlation is due to the low Young’s modulus of UoS-IF-#39, which causes it to compress and extend easily during bending, allowing the portion of the printed structure composed of UoS-IF-#4 to move closer to the neutral axis. The relatively high Young’s modulus of UoS-IF-#4 means that during bending the UoS-IF-#39 layer will deform significantly and reduce the deformation of the UoS-IF-#4 layer in the middle. Taking this into account requires more sophisticated modelling of the materials and structures, and is beyond the scope of this research. Where these practical results could influence the model used in any future work it is noted, but the model from the previous section is not further developed in this work.

It was also noted during this test that there was a difference between the behaviour of printed structures with different widths. This gives support to the hypothesis that a model should account for the fixed width of textile that is not directly beneath the printed structure but must still be extended or compressed during bending of the printed structures.

It was found that as the proportion of UoS-IF-#4 in the beam increased the difference between the mandrel durability of wide (3 mm) and narrow (1.5 mm) printed conductive tracks decreased. The interpretation of this is that with small amounts of UoS-IF-#4 the effective stiffness of the printed structure is low in relation to the extra textile that must be extended, this extra textile being the same for 3 mm wide and 1.5 mm wide structures. As the amount of UoS-IF-#4 increases the effective stiffness of the printed structure increases and the effect of the extra textile is diminished. The results from a 3 mm radius mandrel test of internal bending with the 75/25, 50/50 and 25/75 UoS-IF-#39 structures that have widths of 1.5 and 3 mm are shown in Table 6-7. Three conductive tracks were tested for each layer structure/width configuration.

Table 6-7: Durability of 1.5 and 3 mm wide tracks with three different interface and encapsulation structures.

<table>
<thead>
<tr>
<th>Designation</th>
<th>75/25</th>
<th>50/50</th>
<th>25/75</th>
</tr>
</thead>
<tbody>
<tr>
<td>UoS-IF-#39</td>
<td>25%</td>
<td>50%</td>
<td>75%</td>
</tr>
<tr>
<td>UoS-IF-#4</td>
<td>75%</td>
<td>50%</td>
<td>25%</td>
</tr>
<tr>
<td>Width (mm)</td>
<td>1.5</td>
<td>3</td>
<td>1.5</td>
</tr>
<tr>
<td>ΔR at 200 cycles (R/R₀)</td>
<td>5.31</td>
<td>5.54</td>
<td>5.13</td>
</tr>
<tr>
<td>ΔR₁,₁₉/ΔR₃</td>
<td>0.96</td>
<td>1.16</td>
<td>1.59</td>
</tr>
</tbody>
</table>

This demonstrates the necessity of the textile width in the model being greater than the width of the printed track. The less stiff the track, the more effect the extra textile has on behaviour during flexure.
6.3.2 Optimisation of selected layer structure

This experiment takes the results of previous sections and tries to find an optimal conductive track structure. Previous experiments were performed using printed conductive tracks on a substrate of Escalade textile. This textile was used to provide large resistance changes during experiments which were statistically separable from the error bars. It was theorised that switching to a Lagonda substrate, which has been shown to have a lower thickness and a greater stretchability, would reduce resistance change during experiments. For this final optimisation experiment Lagonda was used. The 50/50 layer structure is assumed to be optimal on Lagonda as it is on Escalade. This assumption is justified in the conclusion of this chapter.

Using the 50/50 layer structure selected in the previous section, a mandrel test was performed to find the optimal placement of the conductive layer in this printed structure. The interface layer is kept to the minimum required thickness in order to minimise the thickness of the printed structure. Three interface and encapsulation structures were printed on to Lagonda with UoS-IF-#4/UoS-IF-#39 ratios 43/57, 50/50 and 55.5/44.5 and printed thicknesses of 250, 285 and 320 µm. These are denoted here as structures 3, 4 and 5. Each structure has two or more possible positions for the silver. The eight different conductive track layer dimensions from the three different interface and encapsulation structures are described in Table 6-8.

Table 6-8: The layout of various interface and encapsulation structures based on structure 50/50. All thicknesses are given in µm.

<table>
<thead>
<tr>
<th>Printing Order</th>
<th>Layer:</th>
<th>Material:</th>
<th>Layer thickness in each structure:</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>3.a</td>
<td>3.b</td>
</tr>
<tr>
<td>1</td>
<td>Interface</td>
<td>#39</td>
<td>70</td>
</tr>
<tr>
<td>2</td>
<td>Conductor</td>
<td>#4</td>
<td>35</td>
</tr>
<tr>
<td>3</td>
<td>Encapsulation</td>
<td>Silver</td>
<td>5</td>
</tr>
<tr>
<td>4</td>
<td></td>
<td>#4</td>
<td>70</td>
</tr>
<tr>
<td>5</td>
<td></td>
<td>#39</td>
<td>70</td>
</tr>
<tr>
<td></td>
<td>Printed Thickness ($t_p$):</td>
<td>250</td>
<td>250</td>
</tr>
<tr>
<td></td>
<td>Total Thickness ($t_t$):</td>
<td>540</td>
<td>540</td>
</tr>
<tr>
<td></td>
<td>Silver position ($h_s$):</td>
<td>105</td>
<td>140</td>
</tr>
</tbody>
</table>

Eight conductive tracks were fabricated for each structure type/silver position combination. Four were tested in each bending orientation using the standard mandrel. 200 cycles were initially performed in internal and external bending. As the rupture of tracks was of greater interest in external bending, the number of cycles was increased beyond 200 until a rupture in the track was observed. The resistance change after 100 mandrel cycles is shown in Figure 6.3-3.
The results show that the resistance increase from internal bending was decreased when the silver layer was closer to the textile in the printed structure. For the thickest cross sections, 5.a-d, there is no clear difference between the durability of 5.a and 5.b, possibly because the neutral axis of structure 5 in internal bending lies between the silver positions 5.a and 5.b. Interface and encapsulation screens with an emulsion thickness lower than 30 µm would be required to confirm this.

For external bending, the resistance change showed little dependence on the structure or the placement of the silver, although for structure 5 a possible minimum of resistance increase can be observed at position b. This was not observed in previous tests but this is the first external bending mandrel test on Lagonda, which has a smoother interface than Escalade and consequently the printed structure should be more uniform and the neutral axis less varied than in printed structures using Escalade as the substrate.

In external bending the textile has a Young’s modulus of zero, so the cross section is varied through the length of the track. Figure 6.3-4 shows how the position of the neutral axis could meander through the length of a conductive track, spreading the strain equally across several potential silver positions. The illustration of structure 5, in Figure 6.3-4 (b), shows how a meandering neutral axis could cause similar small deformations at silver positions 5.b and 5.c and larger deformations at 5.a and 5.b.
The number of external bending cycles before rupture was greater for a thicker structure because it took longer for sufficient internal damage to accumulate and form a rupture through the whole printed structure. The external bending tests were carried on beyond 200 cycles in increments of 100 cycles until the tracks broke. Taking an average of the number of cycles elapsed before a track was broken, the average cycles until break were 212.5, 200 and 262.5 for structures 3, 4 and 5 respectively. The fact that structure 3 has a slightly higher average seemed to be caused by a single anomalous result, except for this anomaly all conductive tracks tested with structures 3 and 4 had ruptured after 200 cycles of external bending.

The ruptures that occur are due to the effect of cyclic stress. The tensile test of UoS-IF-#4 in Section 4.2.5 showed that even at the low extensions encountered during bending at a 3 mm radius there is plastic deformation and consequently it is likely that there will be a work hardening effect in this material, making it more brittle with repeated stresses. This may be due to micro-fractures that form in the material. A stiff material that deforms more elastically would produce conductive tracks that have a greater endurance to cyclic stress.

6.3.3 Washing test with optimised UoS-IF-#4 and #39 structures

A machine washing test was carried out with structures composed of both UoS-IF-#4 and UoS-IF-#39 to find whether the durability had been improved. The test regime was the same as that described in the machine washing test for UoS-IF-#4 conductive track structures, however further steps were taken to ensure the reliability of via connections. The width of the UoS-IF-#39 interface and encapsulation was increased to 4 mm to ensure the UoS-IF-#4 was protected from moisture on the sides as well as above and below. The UoS-IF-#4 interface and encapsulation and the conductive layer had widths of 3 mm and 2 mm respectively.
36 conductive tracks were fabricated using structure 5.b. as described in section 6.3.2. This structure was chosen as it had the lowest resistance increase and the fewest ruptures in mandrel testing. The 36 tracks were printed in three-track 12 patches. These were sewn on to a backing of 50 mm wide Elasta textile. Three patches were sewn on to each of four strips of Elasta. One of these strips of Elasta is shown in Figure 6.3-5.

![Figure 6.3-5: Three printed Lagonda patches sewn on to an Elasta strip backing.](image)

Vias were created by punching a steel button through the printed structure, which was subsequently encapsulated on the printed side with UoS-IF-#39. This was used because of its low water absorption. Stencil printed and glob top encapsulation methods were employed. 18 conductive tracks were encapsulated with each method. These methods were shown previously in section 5.7.2.

The four strips were washed at 40 °C for 59 minutes in a wash cycle with a maximum spin speed of 1000 rpm. They were then dried for 1 hour at 40 °C, and the resistance of each track was measured. Commercially available dissolvable sachets of liquid detergent were used in each wash. The resistances were measured before the first, after the fifth and after the tenth wash. One of the conductive tracks was non-conductive before the test began and another was broken when the conductive tracks’ resistances were measured after the tenth wash. The resistances for the 34 of 36 conductive tracks which were unbroken during the test are shown in Figure 6.3-6.

![Figure 6.3-6: Resistance increase of conductive tracks structure 5.b during the washing test.](image)

The electrical resistance increase after 10 washes was around 200%. While this is high, it would be acceptable for biopotential monitoring applications because a significantly higher resistance (1-1000 kΩ) is inevitably encountered at the electrode and the skin-electrode interface.
The two tracks that showed no conductivity were both encapsulated with the stencil printing method and showed no ruptures along the length of the track, unlike those observed in the first washing test. This suggests that the mode of failure for the conductive tracks that were non-conductive was a problem with the via encapsulation structure, since none of the glob-top encapsulated structures broke. Further, there were several vias encapsulated by stencil printing where the encapsulation broke off from the conductive track during washing, 6 out of 36 in total, as shown in the two via encapsulations at the bottom left of Figure 6.3-7. Most of these remained conductive.

![Figure 6.3-7: Two stencil printed encapsulations having broken off (bottom left) after 10 washes.](image)

This could be due to a low curing energy at the bottom of the stencil printed encapsulation structure, as UV light is absorbed and reflected by polyurethane higher in the structure and closer to the light. This might cause poor bonding between the stencil printed encapsulation meets the screen printed base. The thickness of the stencil and the metal buttons may have reduced the curing energy at various points. Further, the glob-top encapsulated structure was less likely to encounter a force separating it from the screen printed substrate as it had no corners or surfaces normal to the textile.

### 6.3.4 Conclusions

A methodical optimisation process was used to examine the behaviour of printed, composite encapsulation conductive tracks on textiles and discern which structures provided the best durability during mechanical deformation and machine washing. The structure eventually adopted had a printed thickness of 320 µm on a substrate of 290 µm thick Lagonda textile. This structure had a 97.1 % durability rate after ten washing tests, with only 1 out of 35 tracks breaking during the tests. No ruptures were observed on any of the conductive tracks and it is thought that a batch of glob-top encapsulated tracks would have a 100 % success rate; the 18 glob-top encapsulated conductive tracks subjected to machine washing remained conductive throughout the test. The resistance increase of these printed conductive tracks after ten washes was around 200 % with Dupont 5000, however repeating the test with a more durable silver paste could reduce this. Electrodog 725A had only around half the resistance increase of Dupont 5000 when compared with a mandrel test in section 4.3.4.
6.4 Conclusions

A neutral axis which is in the same position for internal and external bending does not exist for printed textile structures because the textile has different characteristics in extension and compression. The textile compresses with minimal stress, due to the air gaps in the structure, but is difficult to extend, meaning that the neutral axis position is closer to the textile side in internal bending than in external bending. For example, the mandrel test results for structure 5 on page 116 demonstrate that the best position for internal bending is 5.a ($h_s = 105 \mu m$). The best position for external bending, however, is around positions 5.b and 5.c ($h_s = 140-175 \mu m$).

A stiff central structure encased in a more elastic material is consequently a sensible approach because the elastic material will compress and the stiffer material will move closer to the neutral axis in either type of bending. If the stiff, inner beam is too thin it will be deformed with too low a stress and if it is too thick there will be no room for the more pliable material to absorb the stress. In this work a stiff beam of UoS-IF-#4 is encased in more pliable layers of UoS-IF-#39.

Between the two textiles examined here, Lagonda provides more durability than Escalade because it has a lower thickness and a finer weave. Samples 50/50 (from section 6.3.1) and 4.b (from section 6.3.2) were identical structures, printed on to Escalade and Lagonda respectively. After 100 cycles of internal bending the former structure had an increase of around 200% in resistance while the latter had only around 20%. The lower thickness of Lagonda causes the whole structure to be thinner overall and reduces the strain on the printed structure. A smooth interface surface can be created in only three prints on Lagonda compared to four on Escalade, allowing the conductive layer to be printed closer to the neutral axis and further reducing its deformation.

Reducing the number of interface deposits required to create a smooth surface would be beneficial to the durability of these flexible conductive tracks. The silver could be printed after just two interface deposits on Lagonda with the existing methods but this interface surface would be rougher, as shown on page 59, and would cause a significant increase in the initial resistance of the conductive track. Further work is required to determine whether the increase in initial resistance would be offset by a sufficient improvement in durability. Reducing the required thickness of the interface layer could involve a decreased screen emulsion thickness, which would mean each deposit has a lower thickness, or an adjustment to the rheology of the paste so that it self-levels more effectively.

This chapter has demonstrated that durable screen printed conductive tracks on woven textiles are achievable. It is likely that future improvements to the materials used for this purpose will further improve durability. It is hoped that the properties of the composite textile printed conductive tracks which have been examined in this chapter, particularly the variation in behaviour observed in internal and external mandrel bending, will aid design processes for this technology in future.
7 Electrode evaluation

7.1 Introduction

This chapter discusses experiments performed to optimise the performance of the electrodes developed in this thesis. The basic design of passive electrodes will be discussed and experiments will be carried out to examine the effect of the dimensions of the electrode and the formulation of the carbon loaded rubber. The effect of such as contact pressure stability and change in performance over time, will also be examined. Then, active electrodes are fabricated using the same technology. The performance of these active electrodes is compared to the passive electrodes developed in this work and commercially available Ag/AgCl electrodes.

7.2 Experimental procedures in this section

Most of the experimental procedures described here are based on the methodology used by Searle and Kirkup [19], who use recordings from electrodes positioned on the arm to compare wet, dry and hydrogel electrodes. In this section recordings are taken with printed electrodes attached to an elastic strip that is worn on the author’s arm. As electrodes can pick up EMG signals from the arm muscles, which can confuse results, steps are taken to ensure that for each recording the arm is relaxed. The signal is also observed during recordings so that muscle EMG signals can be visually identified. Even a slight tension on a muscle near the recording electrodes can result in EMG spikes, with a repetition frequency on the order of 10 Hz but varying depending on which muscle is observed. This effect is shown by observing the signal at different times during a single recording using a differential pair of carbon loaded rubber electrodes on the forearm positioned 7 cm apart on the anterior side of the forearm, 10 cm distal from the elbow pit. During this recording the forearm was first relaxed and resting on a desk, and then the hand is raised into the air. Figure 7.2-1 and Figure 7.2-2 show the signals during the relaxed and raised hand periods respectively. A 2nd order Butterworth high pass filter at 6 Hz is applied to remove any baseline drift.
There is a clear difference between the signals recorded in the two arm positions. EMG noise is picked up from the tensed muscle. This could act as a confounder in the comparison of different electrodes, so the same body position must be maintained in any compared recordings. In most experiments described here, several trials are averaged to improve accuracy and reduce the confounding effect of EMG signals and uncontrollable changes in skin impedance. Where possible, experiments are performed in a single day, to prevent further confounding effects due to longer term changes in skin properties.

This section examines the effect of electrode contact area, electrode resistance, electrode thickness and contact pressure. Settling time and impedance at different frequencies are also investigated. Measures are taken to keep all variables other than the investigated variable constant, as described in Table 7-1.

Table 7-1: The main experiments in this section and the steps taken to keep other variables constant.

<table>
<thead>
<tr>
<th>Section:</th>
<th>Experiment:</th>
<th>Variable:</th>
<th>Contact Area</th>
<th>Electrode Resistance</th>
<th>Contact Pressure</th>
</tr>
</thead>
<tbody>
<tr>
<td>7.3.2</td>
<td>Contact Area</td>
<td>Variable</td>
<td>Constant</td>
<td>Constant</td>
<td>Constant</td>
</tr>
<tr>
<td>7.3.3</td>
<td>Electrode Resistance</td>
<td>Constant</td>
<td>Variable</td>
<td>Constant</td>
<td>Constant</td>
</tr>
<tr>
<td>7.3.5</td>
<td>Contact Pressure</td>
<td>Constant</td>
<td>Constant</td>
<td>Constant</td>
<td>Variable</td>
</tr>
<tr>
<td>Notes</td>
<td></td>
<td>Constant = All electrodes have the same dimensions.</td>
<td>Constant = All electrodes are made with the same carbon formulation.</td>
<td>Constant = All textile supports are the same length and applied to the same part of the body.</td>
<td></td>
</tr>
</tbody>
</table>

Recordings made using “3M RedDot self-adhesive Ag/AgCl resting EKG electrodes”, hereafter referred to as the 3M RedDot electrodes, are used as a reference throughout this section. The procedure for these recordings is described in each case. The electrodes have an Ag/AgCl layer with a 500 µm thick hydrogel layer on top. They have a contact area of 22 x 22 mm and a tab
for attaching clips to connect to the amplifier. The 3M RedDot electrodes are shown in Figure 7.2-3.

![3M RedDot electrodes](image)

Figure 7.2-3: 3M RedDot electrodes with skin contact area 22 x 22 mm.

The experiments described in this section are not clinical trials, and do not serve to prove that these electrodes are appropriate for any specific medical application; investigations into various applications for these electrodes are described in chapter 8. The aim of these experiments is to ascertain whether these electrodes follow previously established rules regarding electrode design and to establish some rules to guide the design of systems using screen and stencil printing on to textiles. All experiments are carried out on the author and further work is necessary to evaluate these electrodes in a wider context but this was beyond the scope of this thesis.

UoS-IF-#4 was used for the interface and encapsulation paste and Dupont 5000 silver paste is used for the conductor for all the experiments described in this chapter. The thicknesses of the interface, conductive and encapsulation layers are 150 µm, 5 µm and 70 µm respectively.

### 7.3 Passive electrodes

A passive electrode is constructed by screen printing an interface layer on the textile. Silver paste is printed on to this interface. A stencil print of carbon loaded rubber is then used to encapsulate the silver layer and electrically connect the top of the electrode, where skin contact occurs, to the bottom of the electrode and the encapsulated conductive track. These electrodes were described previously in section 3.5.2. This section examines various design choices and their effect on the performance of the passive electrodes developed in this work.

#### 7.3.1 Signal acquisition

Signals are acquired by electrically connecting electrodes, attached to the skin surface, to an instrumentation amplifier. The amplifier selected for this work is a bipolar amplifier with a gain of 1000. The amplifier circuit schematic is shown in Figure 7.3-1.
Figure 7.3-1: The amplifier described by Spinelli et al [74].

This amplifier was selected because the AC coupling network (A) reduces baseline drift by coupling AC into the amplifier and dissipating DC charge through the resistors. This is especially useful with electrodes such as those used in this thesis which, unlike Ag/AgCl electrodes, have no electrochemically stable potential at the skin and consequently tend to have significant levels of baseline drift.

Differential electrodes are connected to $E_1$ and $E_2$ and the driven right leg electrode is connected to $E_3$. The functions of the various parts of the amplifier can be described individually.

- A is the AC coupling network, which reduces DC drift.
- B is a normal instrumentation amplifier with a gain of 1000. The gain is defined by the resistors $R_3$ and $R_4$, and is equal to $2R_4/R_3$.
- C is a second order low-pass filter with a cut-off at 150 Hz.
- D is a driven right leg (DRL) circuit which drives the body being examined with an inverted common mode signal to reduce the noise level in the signal.
- E is an integrator which drives the output signal back to 2.5 V. It can effectively be thought of as a high pass filter with a low (<1 Hz) cut-off frequency.

Different amplifiers have different designs, so all measurements described in this section are specific to this amplifier. All measurements are taken with the amplifier in an electrically shielded box and the textile electrode network is connected with shielded cables. Textile printed conductive tracks that form part of the signal path are not shielded. Four of these amplifiers were fabricated for this work.

The amplified signal at $V_0$ is connected to a National Instruments USB-6008 data acquisition card which samples at a rate of 1000 Hz and a 10-bit depth over 10 V. This provides a resolution of 10 mV. The signals are filtered and recorded with National Instruments Labview and any further analysis is performed with MathWorks MATLAB.
7.3.2 Contact area

The first electrode parameter to be examined here is the skin contact area. Huigen et al [75] found that signals taken from differential skin contact electrodes had a relationship between noise, $n$, and contact area, $A$, as described in Equation 3.

\[
E_{\text{3}}: \text{Relationship between electrode contact area and noise as described by Huigen et al [75].}
\]

\[
n \propto \frac{1}{\sqrt{A}}
\]

To examine whether this applies to screen and stencil printed electrode systems, circular electrodes were designed with diameters 10, 20 and 30 mm. The screen design, EL-2, is shown in Figure 7.3-2.

The conductive tracks can be printed very narrow, the narrowest in this work is 500 µm wide. Consequently the majority of the silver paste used in printing these systems goes towards fabricating the electrodes. Given that the noise level is thought to be largely dependent on the skin contact area rather than the silver area, an electrode design was created that had a 20 mm diameter but reduced volume of screen printed silver, denoted design M2. If this shows comparable performance to design M1 then less silver paste can be used in future designs, potentially cutting the silver usage in such a system by 50%.

The formulation used for the electrode paste was the standard carbon loaded rubber formulation and the electrodes had thickness 3 mm. The screen printed part of these
electrodes is shown in Figure 7.3-3 and the electrodes after the stencil printed carbon loaded rubber is added are shown in Figure 7.3-4. Two copies of this set of electrodes were fabricated.

![Figure 7.3-3: Four electrodes with diameter 10, 20 and 30 mm after screen printing.](image)

![Figure 7.3-4: Four electrodes with diameter 10, 20 and 30 mm with stencil printed carbon loaded rubber.](image)

Matching electrodes were placed with the carbon loaded rubber electrodes face to face and the resistance from the via pad of one electrode to the via pad of the other was recorded. Contact pressure where the electrodes contact one another was controlled with the use of weights placed on the electrode, and was maintained at 12.5 kPa for each measurement using small weights. The resulting resistances are given in Table 7-2.

<table>
<thead>
<tr>
<th>Electrode Design:</th>
<th>L</th>
<th>M1</th>
<th>M2</th>
<th>S</th>
</tr>
</thead>
<tbody>
<tr>
<td>Contact Area (cm²):</td>
<td>7.069</td>
<td>3.142</td>
<td>3.142</td>
<td>0.785</td>
</tr>
<tr>
<td>Face-to-Face Resistance (Ω):</td>
<td>4.9</td>
<td>9</td>
<td>12</td>
<td>38</td>
</tr>
</tbody>
</table>

This result demonstrates the advantage of increased contact area and also shows that the electrodes obtained using the standard carbon loaded rubber formulation have very low resistance. These patches were attached to an arm band, via connections were made with steel buttons, and terminal blocks were connected to the steel buttons on the back of the textile to facilitate connection from the electrodes to the amplifier. The front and back of the armband are shown in Figure 7.3-5 and Figure 7.3-6 respectively.
In these figures the vias are not encapsulated with polyurethane. Most wearable networks fabricated for use in biopotential recordings were not machine washed, and so did not need the extra durability. Insulating tape was used instead to cover the via connections and prevent them from making contact with the skin.

Using the printed passive electrodes L, M1, M2 and S arranged in four differential pairs, a differential pair of 3M RedDot Ag/AgCl electrodes and a 3M RedDot Ag/AgCl driven right leg electrode, recordings were taken with the standard biopotential amplification apparatus. Four five-minute recordings were taken from the upper arm of the author over the course of a day. The electrodes were situated on the anterior side of the upper arm to allow easy access to the electrodes, and the Elasta band was fastened around the top of the upper arm. Recording began 20 seconds after the electrodes were attached. A 2\textsuperscript{nd} order Butterworth band-pass filter is applied with lower and upper limits of 40 and 60 Hz respectively and the RMS noise from each electrode pair in each recording is calculated in Labview. The results were averaged and are shown in Figure 7.3-7. Error bars show 1 standard deviation.
The results show that a larger electrode contact area reduces noise. Clearly covering a larger skin area greatly reduces the impedance of the electrodes, and thereby reduces noise caused by impedance mismatch. It is not possible to conclude from these results whether the difference between the design of electrodes M1 and M2 caused a difference in performance due to the relatively high variability in the measurements, resulting in a high standard deviation. All printed electrodes had greater noise than the commercial Ag/AgCl electrodes. Based on Huigen’s work, the noise should be proportional to the square root of the contact area, so if the RMS noise with electrode design S is $k$, design L should have an RMS noise of $0.33k$ and designs M1 and M2 should have an RMS noise of $0.5k$. In this experiment, the RMS noise levels measured for electrode designs L, M1 and M2 were 0.26k, 0.39k and 0.49k, close to the expected values. Design M1 appeared to acquire signals with less noise than M2 although this experiment was insufficiently precise to show with certainty whether M1, which used around twice as much silver paste, provides superior performance. The M2 low silver paste usage electrode design was consequently not adopted. Improvements to this experimental procedure are discussed in the conclusions of this section.

7.3.3 Electrode resistance

The level of noise varies with the impedance between the signal source (heart or muscle) and the amplifier. This impedance includes tissue, skin and the electrode itself. The observed noise level in a signal depends on all components of the impedance. This experiment aims to examine the effect of electrode resistance on the level of noise.

Given that the skin impedance varies greatly with the level of moisture in the skin, Ten20 EEG conductive gel is used in this experiment to minimise the effect of the skin impedance, and maximise the correlation between the electrode resistance and the observed noise. Round electrodes with thickness 1.2 mm and diameter 10 mm are fabricated with four different
carbon loaded rubber formulations, described in Table 7-3. The 20 phr formulation is the standard carbon loaded rubber formulation.

Table 7-3: The resistance of electrodes fabricated with various carbon loadings in the carbon loaded rubber paste. Resistance is measured from the electrode surface to the via pad with the standard multimeter. The average resistance of three electrodes is shown here.

<table>
<thead>
<tr>
<th>Carbon (phr)</th>
<th>Thinner (phr)</th>
<th>Resistance (Ω)</th>
</tr>
</thead>
<tbody>
<tr>
<td>14</td>
<td>10</td>
<td>2483333</td>
</tr>
<tr>
<td>16</td>
<td>10</td>
<td>4333</td>
</tr>
<tr>
<td>18</td>
<td>10</td>
<td>325</td>
</tr>
<tr>
<td>20</td>
<td>10</td>
<td>54.3</td>
</tr>
</tbody>
</table>

A three electrode patch is fabricated using screens PC-2 with each of these formulations. Electrodes are stencil printed with a diameter of 10 mm and a thickness of 3 mm. These are connected to via pads by conductive tracks of length 40 mm and conductor width 2 mm. Each patch is sewn to an elastic strip with hook and loop textile at the ends. The via pads have steel buttons clamped through the textile.

Using the two outer electrodes as differential electrodes and the central electrode as the DRL, four five-minute recordings are taken with each set of electrodes throughout a single day. At the beginning of each recording the electrodes are reattached and have gel reapplied. The electrode construction is shown in Figure 7.3-8 through Figure 7.3-10.

By ensuring all the bands have the same dimensions, and fastening them at a fixed point on the arm in a fixed posture, the contact pressure is kept constant for each trial. The bands are placed around the forearm as described for the recordings on page 122. The average RMS noise from the four readings was calculated in Labview and is shown in Figure 7.3-11.
The noise level increases with electrode resistance, as does the variability of the noise level. The more carbon added to the electrode formulation, the less noise is expected when the electrodes are used in practice. The level of noise depends on the imbalance in impedance between two electrodes, the magnitude of which is likely to increase as the material resistivity increases. Although there is a clear correlation between carbon loading and noise in these results, there is a high level of standard deviation in these results, possibly due to differences in the amount of conductive gel used or skin preparation.

### 7.3.4 Settling Time

The settling time effect refers to the gradual decrease in noise in the acquired signal over the first few minutes after an electrode is applied. The settling time effect occurs due to the skin reacting to the electrode placed on it. The most commonly cited explanation for this is that the skin heats up, due to being enclosed by an electrode, and consequently sweat pores open to cool the skin [76]. Using the results from the experiments described in section 7.3.2 (contact area) and section 7.3.3 (resistance) two graphs are plotted. The graphs, shown in Figure 7.3-12 and Figure 7.3-13, show the RMS noise over 5 minutes for the best and worst performing electrodes from each experiment. These are the largest and smallest electrodes from the contact area experiment and the 14 phr and 20 phr electrodes for the conductivity experiment. An average of all recordings is also shown.
In both cases the level of 50 Hz noise falls to almost half the initial level after 5 minutes. In the electrode contact area experiment, the fall in noise to a settled level appears to be sharper, with a larger difference in noise between the first and second minutes. The main difference between these experiments was that conductive gel was used in the experiment examining electrode formulation. This may suggest that conductive gel prevents the skin from responding as quickly to being covered. However, there is an equally sharp fall in noise after the first minute for the 20 phr electrodes in the contact area experiment, so it may be the case that the use of a more conductive carbon formulation reduces the length of settling time for these electrodes as well as the overall noise.

7.3.5 Contact pressure

Some research has suggested that the level of contact pressure affects the impedance at the skin electrode interface [77]. To investigate this, recordings were made using printed electrodes resting on the arm and printed electrodes held down with a 1 kg weight. For these electrodes it was found that there was little correlation between the contact pressure and the level of noise and motion artefact if the contact pressure was kept constant.

However, a change of pressure at one of a pair of differential electrodes caused a significant motion artefact on the signal. Figure 7.3-14 shows a recording taken from a pair of circular printed electrodes with a 20 phr carbon loading a diameter of 10 mm. Pressure is applied alternately to one electrode then the other using a plastic rod. Green dotted lines indicate pressure applied to the electrode connected to input terminal E1, and red dotted lines indicate pressure applied to the electrode connected to input terminal E2.
This suggests that contact pressure does not make a significant difference to the skin-electrode impedance for these electrodes but does affect DC behaviour and causes baseline drift, as indicated previously in experiments in section 7.3.5. Investigation into the impedance level during these measurements using a bio-impedance setup would shed further light on these mechanisms. A further investigation, using the same setup as described in Figure 7.3-14, involves the application of pressure to one of the electrodes, waiting for the amplifier to stabilise at 2.5V, and then releasing the pressure. The unfiltered recording is shown in Figure 7.3-15.

The noise level does not change observably when this contact pressure is applied, suggesting that the impedance mismatch between the electrodes is unaffected. The release of pressure causes a DC change in the opposite direction to the change caused by the application of pressure. This problem does not occur to the same extent with Ag/AgCl electrodes, as they electrochemically stabilise the potential at the electrode so that these DC changes are not seen.
at the amplifier. Figure 7.3-16 repeats the experiment described in Figure 7.3-14 with commercial Ag/AgCl electrodes rather than printed textile electrodes.

![Figure 7.3-16: Recording as pressure is applied alternately to each electrode in a differential pair of commercial Ag/AgCl electrodes placed on the forearm.](image)

Compared to Figure 7.3-14 the amount of baseline drift resulting from the applied pressure is greatly reduced because the DC potential at the skin is stabilised electrochemically. Although comparable performance in terms of susceptibility to AC noise can be achieved, the stability of the potential with the printed electrodes is poor in comparison to Ag/AgCl and consequently baseline drift will be present in electrodes of this design. This can be improved by taking steps to stabilise the contact pressure. Using a textile band to hold the electrodes in place spreads any pressure applied, and a differential pair of electrodes that are physically close together on the skin are partially immune to this effect as pressure tends to change equally on both electrodes. Other ways of stabilising pressure with textile electrodes have been proposed in the literature; for example Ottenbacher et al [78] use foam beneath woven electrodes to minimise pressure changes. This method is explored during the fabrication of a Frank configuration vest in section 8.3.

Although the 50 Hz interference can be minimised by paste formulation and electrode design, the susceptibility to motion artefact and loss of contact is a problem in dry electrodes. Commercially available disposable electrodes usually have an adhesive gel, which provides a stable contact with the skin. This is not possible in the designs in this thesis, as the gel degrades over time and consequently an electrode using this technology will not be durable. Consequently, the electrodes are held to the skin by an elastic textile which exerts pressure on the skin via the electrode.

The effectiveness of using stencil printed electrodes to increase electrode thickness and make a more stable contact is evaluated using two ECG bands, one with screen and one with stencil printed electrodes. The electrodes for these bands is screen printed using screens EL-1. These bands have a fixed length so that the same strain occurs on each band when it is worn. Several motions are performed by the author while wearing each of these bands to examine their susceptibility to the artefacts. The two bands are shown in Figure 7.3-17.
Figure 7.3-17: Screen printed (top) and stencil printed (bottom) carbon loaded rubber electrodes printed on identical chest bands for ECG monitoring.

The two electrodes on each band are used as differential electrodes and a 3M RedDot electrode is used as the DRL. The band is worn so that the differential electrodes are on opposite sides of the chest, equidistant from the centre of the body and 25 cm inferior from the top of the shoulder. The electrode spacing is 30 cm across the skin. After a five minute settling period, a four minute series of movement tasks were carried out.

The tasks were split into a typical arm movement task, an extreme arm movement task and a posture change task. The typical arm movement task involved moving the right arm out perpendicular to the body. The extreme arm movement task involved moving the arms above the head, parallel with the line of the body. The posture change task involved bending over to pick up an object on the floor. Each experiment is carried out twice, once with 5 mm thick stencil printed electrodes and once with 60 µm thick screen printed electrodes. The standard carbon loaded rubber formulation was used for both sets of electrodes. The bands are identical apart from the fabrication of the electrode.

**Typical arm movements:**

The typical arm movement recordings showed minimal disruption to the ECG signal. There was a small amount of EMG noise from the torso muscles that are contracting to maintain this posture. The baseline drift was very low for the screen printed electrodes, but changes in pressure caused some baseline drift with the stencil printed electrodes. Figure 7.3-18 and Figure 7.3-19 show segments of the ECG recorded with the chest bands as the right arm is held out to the side with the screen printed and stencil printed electrode chest bands respectively.
Figure 7.3-18: An ECG recorded with screen printed electrodes as the right arm is lifted out to the side, held for five seconds, and then relaxed by the side of the body.

Figure 7.3-19: An ECG recorded with stencil printed electrodes as the right arm is lifted out to the side, held for five seconds, and then relaxed by the side of the body.

In this recording the stencil printed electrodes had more baseline drift. However, there was no signal saturation and the components of the ECG are visible in both recordings.

**Extreme arm movements:**

Figure 7.3-20 and Figure 7.3-21 show segments of the ECG recorded with the chest bands as the left arm is held above the head with the screen printed and stencil printed electrodes respectively.
Figure 7.3-20: An ECG recorded with screen printed electrodes as the left arm is lifted above the head, held for five seconds and then relaxed by the side of the body. The arrow indicates an artefact caused by loss of contact.

Figure 7.3-21: An ECG recorded with stencil printed electrodes as the left arm is lifted above the head, held for five seconds and then relaxed by the side of the body.

The extreme arm movement recordings had increased levels of EMG noise and baseline drift, corresponding with stronger muscle contractions for the more difficult motion and more dimensional change in the torso as these muscles shift around. There was evidence of a loss of contact during the movement in the screen printed electrodes. Figure 7.3-20 shows a clear loss of contact when the movement is performed, indicated with an arrow. Stencil printed electrodes perform better during this task.

**Posture change task:**

Figure 7.3-22 and Figure 7.3-23 show segments of the ECG recorded as the right hand is pressed to the floor from a standing position with the screen printed and stencil printed electrodes respectively.
Figure 7.3-22: An ECG recorded with screen printed electrodes as the right hand is touched to the floor for five seconds and then the body returns to a standing position.

Figure 7.3-23: An ECG recorded with stencil printed electrodes as the right hand is touched to the floor for five seconds and then the body returns to a standing position.

There was significant baseline drift during the posture change task for both sets of electrodes. However, the baseline drift was far greater on the screen printed electrodes, and there is also a clear saturation of this signal between 1.5 and 2 seconds, suggesting that electrode-skin contact was lost.

In conclusion, these recordings suggest that stencil printed electrodes are less likely to lose contact with the skin. However, there are still some motion artefacts in the stencil printed design due to changes in the contact pressure, which alter the potential at the skin-electrode interface and cause DC drift.

7.3.6 Impedance-Frequency Sweep

An impedance-frequency sweep with 400 points from 20 Hz to 10 MHz was performed using two round printed textile electrodes with thickness 3 mm and diameter 10 mm, without conductive paste, placed face to face with a contact pressure of 25 kPa applied using small weights. The electrodes were fabricated with the standard carbon loaded rubber formulation. This measurement was performed with a Wayne Kerr 6500B precision impedance analyser. The impedance and phase angle are shown in Figure 7.3-24.
The DC impedance is around 1 kΩ and remains relatively stable from 20 to 1000 Hz, changing less than 20 %. The phase angle is near 0 at these frequencies. At frequencies higher than 1000 Hz the impedance drops considerably and a negative phase is introduced, indicating a capacitance. Human biopotentials are typically not observed at these frequencies [19] so this is unlikely to affect the performance of the electrodes.

### 7.3.7 Conclusions

The formulation of carbon loaded rubber and the contact area have been examined in order to find design rules to reduce 50 Hz interference. A high conductivity and a large contact area are preferable for low levels of 50 Hz noise. The effect of contact stability has also been examined and it was found that the stability of the pressure rather than the magnitude dictates the level of baseline drift. Stencil printed electrodes perform better than screen printed electrodes during strenuous movements.

Although the noise and motion artefact can be reduced on these passive electrodes, their performance is limited by the high impedance of the skin. As they are dry electrodes, it is not possible to create the ion conductivity necessary to create a stable electrode potential with an Ag/AgCl layer. This means changes in skin-electrode impedance, which can occur for various reasons, will cause noise and baseline drift on these electrodes. Active electrodes are required to reduce the problems caused by high skin impedance.

### 7.4 Active electrodes

Active electrodes work by converting the high impedance signal that occurs at the electrode to a low impedance signal. This is necessary because the impedance of the signal affects the voltages seen at the amplifier inputs. When the signal impedance is high the signal seen by the amplifier is reduced, causing noise that would otherwise be common mode to be observable on the signal. A high impedance signal is also more susceptible to the capacitive coupling of noise in the cables.

Active electrodes buffer the signal from a sensing element, increasing the current at the output while keeping the output voltage the same as the input voltage. This occurs at the electrode so a low impedance signal is transferred through the cable to the amplifier. The
advantages of an active electrode are reduced 50 Hz noise and reduced baseline drift, while the disadvantages are the increased power consumption, the addition of ground and +5 V lines to each electrode, and the added fabrication complexity required to integrate electronic components into the electrode.

7.4.1 Signal acquisition

The front-end amplifier is slightly altered from the standard biopotential amplification apparatus that has been previously described. As the signal from the active electrodes is buffered, there is a high current from these electrodes into the amplification circuit. As the normal amplification circuit has an AC-coupling circuit at the electrode inputs, the signal is fed directly into a capacitor. The relatively high current output from the active electrodes can charge these capacitors, causing instability and preventing the amplifier circuit from operating correctly.

For this reason, the front-end amplification circuit used in these tests has another op-amp stage added before the amplifier. This allows the current to dissipate through the feedback resistor and prevents the capacitors from charging. This circuit can be used with passive electrodes as well, so this circuit is used for all tests described in this section (7.4). The new front-end amplification circuit with the added amplification stage (marked A) is shown in Figure 7.4-1.

![Figure 7.4-1: The amplification circuit used for active electrodes. Box A shows the amplification stage that has been added.](image)

The added amplification stage A has a gain of \( R_x/R_y \). \( R_x \) is 800 kΩ and \( R_y \) is 100 kΩ so the first stage has a gain of 8. The gain of the original amplification circuit is reduced to 100 by altering the values of \( R_3 \) and \( R_4 \) in box B, so this amplifier has a total gain of 800. Using this amplifier with the active electrodes significantly decreases baseline drift compared with the circuit described earlier. Other than the change of gain and the added amplification stage this circuit is identical to that described by Spinelli et al [74].
7.4.2 Experimental procedures in this section

In the previous section, in which passive electrodes were evaluated, the influence of various design factors on the performance of the electrodes was discussed. However, most of the factors examined ultimately affect the impedance of the electrode or the electrode-skin interface. Because active electrodes control the impedance of the signal seen by the amplifier, and because active electrode prototypes are more time-consuming to fabricate, it was considered unproductive to examine contact area, electrode paste formulation and other factors individually and instead this section focusses on a comparison of commercial Ag/AgCl electrodes, printed passive electrodes and printed active electrodes in a Holter ECG monitoring application. As far as possible, the electrode contact area was kept constant with each of the three types of electrodes. The printed passive and active electrodes were fabricated using the same stencil and both have a skin contact area of 23 x 28 mm and the 3M Reddot electrodes have a diameter of 22 x 22 mm. A 3M Reddot electrode was also used as a DRL for all three electrode sets. The printed passive electrodes are fabricated using screen EL-1 and the printed active electrodes are fabricated using the screen AE-1. The three sets of electrodes used in this experiment are shown in Figure 7.4-2 through Figure 7.4-4.

![Figure 7.4-2: Active electrodes used in this experiment. The differential electrodes are on the far left and right.](image)

![Figure 7.4-3: Passive electrodes used in this experiment. The differential electrodes are on the far left and right.](image)
The electrodes were worn around the chest, in the case of the printed textile electrodes, or stuck to the chest with tape in the case of the Ag/AgCl electrodes. The electrode positions and separation (300 mm) were the same in each instance and no conductive gel was used, in order to maintain the same test conditions for each electrode set. One set of electrodes was worn at a time. The subject stood on a treadmill wearing a set of electrodes and successive 2 minute recordings were taken at ambulatory speeds of 0 km/h, 4 km/h, 8 km/h and 12 km/h. The electrodes are worn constantly throughout this recording process, which is repeated for each set of electrodes. There was a break of 30 minutes between the trials of each electrode type.

7.4.3 Data processing

In this experiment a methodology was employed to give two statistical measures, one related to the 50 Hz noise and the other related to baseline drift in these recordings. A measure of baseline drift is calculated by manually detecting and removing the QRS peaks, replacing them with straight lines connecting the remaining data points, then calculating the range and standard deviation of the modified data. The range and standard deviation of the signal can both be used as measures of baseline drift. Figure 7.4-5 (a) shows 4 seconds of a 2 minute ECG recording taken with the printed active electrodes. The range and standard deviation of this recording is shown in Figure 7.4-5 (b). The shaded areas indicate the 100 ms periods from which the QRS peaks were removed.

The 50 Hz noise measure is calculated by filtering this modified data through a 2\textsuperscript{nd} order Butterworth band-pass filter with lower and upper limits at 40 and 60 Hz. The RMS of the resulting data is then calculated to give a measure of the 50 Hz noise present in a signal. The result of this filtering process on the same 4-second recording shown in Figure 7.4-5 (a) is shown in Figure 7.4-5 (c).
7.4.4 Experimental results

The processing of the recordings provides statistics that indicate the noise and baseline drift levels that are encountered during the use of each type of electrode. The standard deviations from the average of the three electrode sets across all movement speeds are shown in Figure 7.4-6.
At low speeds the printed active electrodes perform best, having less baseline drift than the Ag/AgCl or the printed passive electrodes. The passive electrodes have significant baseline drift due to impedance mismatch on the electrodes, but the increase in baseline drift as speed increases is not as significant. The performance of the passive electrodes improves as the skin perspires and its impedance falls. The Ag/AgCl electrodes had to be reapplied after the 8 km/h trial as perspiration caused them to detach. The reattachment may account for the relatively low standard deviation with the Ag/AgCl electrodes at 12 km/h. The 50 Hz noise measure of each electrode set across all speeds is shown in Figure 7.4-7.

This measure is clearly contaminated by motion artefact, as the level of 50 Hz noise should be relatively constant for each electrode type over the recording period. The RMS level for the printed active and Ag/AgCl electrodes at 0 and 4 km/h are very low, and no 50 Hz noise can be observed in these recordings, although some appears after filtering because the components of an ECG combined with muscle and ADC noise cover a large range of frequencies. The 50 Hz
noise RMS levels for these recordings were at least as low as these results suggest, which suggests a maximum peak to peak noise amplitude of approximately 10 mV. The smallest deflection on the author’s ECG, the U deflection, has an amplitude of around 50 mV using the same equipment.

50 Hz noise is visible on the recordings from the passive electrodes at these speeds because the electrode impedance is not well matched. The noise with the passive electrodes does not increase as significantly with ambulatory speed, possibly because perspiration reduces the skin impedance and thereby reduces the magnitude of the impedance mismatch. Another factor that might affect the passive electrode RMS noise measure as ambulatory speed increases is periods of signal saturation. This has very low RMS amplitude after filtering, despite the fact that these signals provide no useful information. Overall, the printed active and Ag/AgCl electrodes had very similar RMS noise levels after filtering during this trial. A summary of statistics from these recordings is given in Table 7-4.

Table 7-4: Standard deviation, RMS noise and range of recorded ECGs with QRS peaks removed for each electrode type at each speed. All measurements are in Volts.

<table>
<thead>
<tr>
<th>Speed (km/h)</th>
<th>Printed Electrodes - Active</th>
<th>Printed Electrodes - Passive</th>
<th>Ag/AgCl Adhesive Electrodes</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>S.D.</td>
<td>RMS</td>
<td>S.D.</td>
</tr>
<tr>
<td>0 km/h</td>
<td>0.04</td>
<td>0.0033</td>
<td>0.19</td>
</tr>
<tr>
<td>4 km/h</td>
<td>0.09</td>
<td>0.0034</td>
<td>0.49</td>
</tr>
<tr>
<td>8 km/h</td>
<td>0.26</td>
<td>0.0136</td>
<td>0.59</td>
</tr>
<tr>
<td>12 km/h</td>
<td>0.53</td>
<td>0.0240</td>
<td>0.65</td>
</tr>
</tbody>
</table>

A recording was also taken at 0 km/h using the active electrodes with an integrated printed carbon loaded rubber passive electrode as the DRL, rather than a self-adhesive Ag/AgCl electrode. This was printed on the textile band, in the remaining electrode pad shown in with the same thickness as the other electrodes. The standard deviation and RMS values from these recordings were slightly higher than with the Ag/AgCl DRL electrode, at 0.066 and 0.0038 respectively. It is thought that this is due to the reduced distance between the differential electrodes and the printed DRL electrode on the textile band, compared to the distance between the textile band and an electrode on the abdomen. These values are still significantly lower than those recorded with the passive electrodes at the same speed.

7.4.5 Conclusions

The active electrodes perform as well as the Ag/AgCl electrodes when the subject is stationary (0 km/h) or walking (4 km/h). Both types outperform the passive electrodes at these speeds. As the speeds increase, the active electrodes have greater susceptibility to motion artefacts than the Ag/AgCl electrodes, with a 71% higher standard deviation at 12 km/h. This is due to the different methods of attachment to the skin, self-adhesive and elasticated. As the motions become more vigorous, the electrodes attached with elastic shift across the skin more resulting in motion artefacts on the amplified signal. This problem does not occur with self-adhesive electrodes. Improving the stretchability of the textile garment would prevent the electrodes shifting so much during movement and reduce the magnitude of motion artefacts.

Overall, these results indicate that the active electrodes have greater baseline stability than the Ag/AgCl electrodes at speeds below 12 km/h but are slightly more susceptible to 50 Hz
noise at 0 km/h, where the effects of motion artefacts on the RMS measurement should be minimal. However, no 50 Hz noise is observable on the active electrodes’ output and the small difference in RMS noise between active and Ag/AgCl electrodes at 0 km/h, 0.8 mV after amplification, may be due to other factors such as coughing during the trial.

These trials are a proof of concept for the active electrode design and fabrication processes described in this thesis. They also allow practicalities, such as the effect of the mounting of the electrodes, to be examined in an application setting. They are not to be viewed as clinical trials as all experiments were performed on a single individual and are subject to any random changes in skin impedance during the trial. More detailed trials, however, are not beneficial at this stage because changes to the garment on which the electrodes are printed will significantly change the motion artefact results.

7.5 Electrode evaluation conclusions

This chapter examined the performance of passive and active electrodes on textiles fabricated with screen and stencil printing technology. The passive electrodes have low levels of 50 Hz noise if their conductivity is high and their contact area is large. Motion artefacts are a greater problem because of DC drift at the electrodes, which is compounded when there are pressure changes. Using thick stencil printed electrodes rather than thin screen printed electrodes reduces the magnitude of motion artefacts but they are still significant during vigorous motions. The printed passive electrodes had good performance compared to Ag/AgCl electrodes at high ambulatory speeds. It is thought that this is due to a significant reduction in skin impedance due to perspiration.

Active electrodes were designed to provide electrodes with improved performance. These were fabricated using the same screen and stencil printing technology as used for the passive electrodes, with the addition of an op-amp and passive components. Tests showed that these electrodes had similar performance to standard commercial Ag/AgCl in terms of rejection of noise, as far as is identifiable with the equipment used here. However, because the printed active electrodes are held to the chest by elastic rather than taped to the chest like the Ag/AgCl electrodes, the pressure changes on the active electrodes during motion were higher. Consequently there were significant motion artefacts on the active electrode recordings, although less than the printed passive electrodes at all tested ambulatory speeds.

This section has demonstrated that electrodes providing good signal quality can be fabricated using screen and stencil printing on textiles. These electrodes are more comfortable, easier to set up and more durable than those currently used for clinical biopotential monitoring. Durability testing in terms of washability has not been carried out on the electrodes in this work, as they require further development to reduce problems with loss of adhesion between the stencil printed carbon loaded rubber and the screen printed base during washing. The adhesion between the carbon loaded rubber and the silver paste layer is sufficient for the electrodes to remain attached during the use of these systems, but it is unlikely that they would survive machine washing. Different methods to firmly fix the carbon loaded rubber electrode to the screen printed base have been proposed in appendix D.
8 Printed wearable monitoring systems

8.1 Introduction

Four wearable electrode networks have been implemented to demonstrate the technology developed in this thesis and to examine the appropriateness of the technology in a variety of human biopotential sensing applications. This chapter has a section focused on each wearable network in the order described below.

As ECG monitoring is the primary focus of this work two wearable ECG electrode networks have been implemented. These are an active electrode chest band for simple, unobtrusive one-lead ECG recording and a Frank configuration monitoring garment in the form of an elasticated vest. Two further systems have been implemented to examine whether the technology is also viable for EMG monitoring, which is used in research, rehabilitation and prosthetic control. These are a textile headband system to record facial EMG and EOG signals, and an upper arm textile band system to investigate EMG signal strength and distribution during the contraction of the bicep muscles. A textile system for functional electrical stimulation (FES) based on work in this thesis is also described. The magnitude of the biopotentials being observed vary, with the largest being the QRS peak in the ECG which has a magnitude of around 1 mV on the skin surface. Smaller ECG deflections and EMG readings with amplitudes around 0.1 mV on the skin surface are also observed. EOG biopotentials have a magnitude of around 0.3 mV on the skin surface.

UoS-IF-#4 was used for the interface and encapsulation paste and Dupont 5000 silver paste is used for the conductor for all the experiments described in this chapter. The thicknesses of the interface, conductive and encapsulation layers are 150 µm, 5 µm and 70 µm respectively unless stated otherwise.

8.2 Active electrode chest band for bipolar ECG

8.2.1 Introduction

This section describes the design, fabrication and testing of a three electrode chest band for monitoring a bipolar one lead ECG. This chest band is aimed at long term Holter monitoring to provide a solution that is more comfortable and durable than existing solutions using Ag/AgCl self-adhesive electrodes that must be replaced daily. By implementing active electrodes this system can achieve this without any reduction in signal quality or diagnostic usefulness compared with existing solutions.

8.2.2 Goals of the system

This textile network was designed to provide a convenient interface for a one-lead, long term ECG. The system uses two differential active electrodes and a passive DRL electrode printed
across a band. This band can be wrapped around the chest. The electrode network on the chest band can be connected to table-top electronics, as they are in this project, or portable electronics, to allow it to be used for remote monitoring.

8.2.3 Layout of active electrode chest band

The differential active electrodes are around 30 cm from each other, on opposite sides of the chest. Using a large separation of the differential electrodes ensures that there will be a significant potential difference wherever the band is worn, and it need not be positioned by a person with medical knowledge to provide an ECG signal. The passive DRL electrode and the vias are positioned between the two differential electrodes. The electrode layout is shown in Figure 8.2-1.

![Figure 8.2-1: The active electrode chest band layout.](image)

The total length of the printed network is around 300 mm, meaning it does not fit on to a single screen, with maximum printed dimensions 200 x 200 mm. Consequently the design shown in Figure 8.2-1 is printed as two separate layouts that are attached to the same piece of textile in the fabrication process. These two layouts are printed using the set of screens AE-1 shown in Figure 8.2-2.

![Figure 8.2-2](image)
Figure 8.2-2: Screen designs AE-1 used for printing active electrodes. (a) is the interface layer, (b) is the conductive layer and (c) is the encapsulation layer.

By keeping the printed conductive tracks between the differential electrodes and signal vias the same length, the 50 Hz noise picked up by both conductive tracks from differential electrodes should be approximately equal in magnitude. This will consequently be a common mode signal and be rejected by the instrumentation amplifier.

8.2.4 Fabrication of active electrode chest band

The chest band structure is fabricated by screen printing the two layouts on to separate 150 x 150 mm squares of Lagonda textile. These two layouts are cut out and sewn on to a 50 mm wide strip of Elasta textile. The active electrodes are fabricated in the manner described in section 5.5. There are seven printed via connections that are used for +5 V, GND, differential electrodes 1 and 2 and the driven right leg. +5 V and GND vias were required for each active electrode as the printed designs are not connected. It would be possible to revise this design to have only five vias if larger screens were used and the full chest band layout could be printed on to a single piece of textile. A steel button was clamped through the textiles to create each via. Wires were wrapped around and soldered to these buttons to ensure a stable electrical contact between the via and cabling connecting to external electronics. An active electrode chest band is shown in Figure 8.2-3
The Elasta strip used has a length of 100 cm. As with all prototypes in this thesis, hook and loop textile is used at the ends of the strip to fasten the band around the body.

8.2.5 Recorded signals from active electrode chest band

The active electrode chest band provides high signal quality without skin preparation. There is no 50 Hz oscillation visible on the obtained readings, meaning that the magnitude of this noise was lower than the resolution of the ADC, 10 mV. An ECG obtained with this wearable electrode network from the author in a seated position is shown in Figure 8.2-4. This recording is has no digital filtering. The bandwidth of the buffer amplifier is 1 MHz so the only filtering on this signal is the amplifier’s integrated low pass filter at 150 Hz, which does not affect the level of 50 Hz noise.

This ECG trace has less noise and baseline drift than traces from passive dry rubber electrodes both in this thesis and in the literature. A full examination of the performance of this belt structure when walking and running has been given in section 7. The performance of the chest band is affected in these scenarios but not to a significantly greater extent than Ag/AgCl self-adhesive electrodes. This active electrode chest band provides low noise and low baseline drift without requiring any adhesive gel, conductive electrode paste or settling time, or causing any observable degradation to the electrodes. This makes them ideal for long term monitoring applications.
8.2.6 Conclusion

There is relatively little dimensional change of this chest band during breathing and walking. During jogging, however, the signal can suffer due to dimensional change, which is less of a problem with self-adhesive electrodes. However, the ECG peaks can be observed even during jogging and with an increased low pass filter frequency in the amplification circuit, which achieved by modifying the integrator portion of this circuit, it is likely that arrhythmia could be picked up with some accuracy during exercise using this system. This prototype demonstrates that this technology could be used for ambulatory one-lead monitoring. The advantage of this technology is that it can be reused and is consequently cheaper than adhesive electrodes that must be periodically replaced. The durability of this device has not been investigated, but as there is a low level of adhesion between the carbon loaded rubber and electrode surface, it is considered unlikely that this version of the chest band is durable to machine washing, however the encapsulation of the circuit should provide high durability if this adhesion problem can be solved.

The other three monitoring systems discussed in this section do not use active electrodes. There is no technical barrier that prevents the described active electrodes being used in these systems but due to the chronology of developments in this work, active electrodes are only used for this prototype system.

8.3 Frank configuration vest

8.3.1 Introduction

A Frank configuration electrode network requires electrodes in various positions around the torso to provide a signal describing the electrical potential across the heart in three different orientations, as described on page 8. A Frank configuration vest was fabricated to demonstrate the applicability of the textile screen printing fabrication process to larger designs. All the other wearable systems described in this thesis are on narrow textile bands with widths of 50-100 mm, so the implementation of this vest system demonstrates that this fabrication process is viable for larger garments, such as the systems created in the MaglC project described previously.

8.3.2 Goals of the system

One goal of this project was to produce a garment to monitor a Frank configuration vectorcardiogram, as was defined in the BRAVEHEALTH project. This uses eight electrodes arranged through a resistive network into three bipolar channels and a ground to give a three dimensional view of electrical activity in the heart. This means the placement of the electrodes is crucial, the heart must be viewed by three separate channels in the X, Y and Z directions. This is achieved by connecting seven electrodes through a passive resistor network. Frank’s original electrode positions were modified for the BRAVEHEALTH project are shown in Figure 8.3-1.
8.3.3 Layout of Frank configuration vest

Slight modifications were made to Frank’s electrode placement for the BRAVEHEALTH project, because it is not possible to position an electrode on the foot using a vest. This foot electrode (F) is consequently moved from the left ankle to the left hip. The BRAVEHEALTH positions were further modified during the final design of the vest; the neck electrode (H) was moved to a position just above the heart. Consequently, the vertically aligned electrodes, providing the y-axis view of the heart, are placed just 30 cm apart, above and below the heart. Frank suggests in his original paper that the positions of the two electrodes that were moved are not as critical as the positions of the X and Z axis electrodes. He described the position of electrode H as “not especially critical” and electrode F as “least critical of all”. Figure 8.3-2 shows a 2D schematic of the vest design.
In order to print the electrodes as shown in this diagram, a large printing area was required. The total area covered by the system was 810 x 330 mm, which is too large for the DEK 248 printing equipment, which could provide a maximum printing area of 500 x 500 mm using large screens. It was possible to print the garment with two separate designs on the same set of 500 x 500 mm screens (FC-1). The interface layer and conductive layer screens from set FC-1 are shown in Figure 8.3-3.

This design could be printed, cut out in two pieces and sewn on to a vest in the configuration shown in Figure 8.3-2. The electrodes shaded in red in this figure correspond to one printed part and green to the other. The electrodes are labelled with letters corresponding with Frank’s design (Figure 8.3-1).

8.3.4 Fabrication of Frank configuration vest
Due to the large area of the print a new substrate holder was required. This increased the printing area to 500 x 500 mm and could be removed from the screen printer for subsequent heat or UV curing of the printed layer. It is supported by four cylinders at the corners and two cylinders placed more centrally to prevent warping due to the printing pressure. This substrate holder is shown in Figure 8.3-4.

Figure 8.3-4: Substrate holder for large area prints.

The fabrication process involved gluing Lagonda to the substrate holder using weak repositional adhesive (3M spramount). The Frank configuration pattern was then screen printed on to Lagonda using UoS-IF-#4 and Dupont 5000 for the interface/encapsulation and conductive layers. As the substrate holder was too large to fit in either the UV cabinet or box oven usually used for curing pastes, a small handheld UV emitter pen with approximately the same output intensity of the UV cabinet was used for the UV curable interface and encapsulation layers. The thermal curing of the conductive layer was carried out using a hot plate heated to 100 °C. This does not fully cure the paste but dries it for the printing of subsequent layers. After printing, the screen printed textile was removed from the substrate holder by heating on the hot plate at 80 °C for 10 minutes which was sufficient to reduce the adhesion of the glue. The two textile parts that make up the Frank configuration design are cut out from this textile and cured in a box oven at 120 °C for 10 minutes to fully cure the conductive paste. The samples after removal and curing are shown in Figure 8.3-5, along with the resistance of each conductive track.
The carbon loaded rubber electrode layer is then stencil printed with the standard carbon loaded rubber formulation on to the square electrode sites with a thickness of 5 mm. Given the size of the samples, making a large stencil to cover all electrodes simultaneously is not cost effective or practical, so each electrode is stencil printed and cured individually. The printed patches after stencil printing are shown in Figure 8.3-6. The measured resistance of each electrode and track is shown. The resistance is measured from the centre of the top of the electrode to the associated via.
The resistances of the conductive tracks are significantly less than the resistances of the electrodes so in terms of impedance these electrodes should behave no differently from previously fabricated passive electrodes despite the increased connection distances.

These patches are then sewn on to a vest. This purpose-built garment is composed of a Lagonda vest cut-out, the printed Lagonda patches and Elasta strips to ensure there is stable pressure on the electrodes. The Elasta strips have hook and loop textile sewn to the end to allow the garment to be adjusted easily. The Elasta is sewn to the outside of the vest and the printed Lagonda textile is sewn to the inside, allowing the electrodes to make contact with the skin. The vest with all textiles sewn together is shown in Figure 8.3-7.

Figure 8.3-7: Frank configuration vest.

Silicone foam of thickness 6.3 mm supplied by eFoam was cut to appropriate shapes and placed between the printed Lagonda patches and the unprinted Lagonda vest at the sites of the electrodes to improve the stability of contact pressure. This appeared to reduce baseline drift although this improvement has not been confirmed experimentally. The electrode structure is shown in Figure 8.3-8.

Figure 8.3-8: Electrode structure used in the fabrication of the Frank configuration vest.

The textile network can then be worn, with all the electrodes making contact with the appropriate parts of the torso. The DC resistance from each electrode surface to the central connection point is less than 1 kΩ, well within the required resistance of a dry electrode as
defined on page 42. The vest is comfortable in a standing or seated position but is less comfortable during movement due to the low stretchability of the majority of the network. The network is worn inside-out in Figure 8.3-9 and Figure 8.3-10 so that the position of the electrodes can be seen.

Figure 8.3-9: Frank configuration vest worn inside out to show the electrode positions; viewed from 45°.

Figure 8.3-10: Frank configuration vest worn inside out to show the electrode positions; viewed from 225°.

Steel buttons are then inserted into the round screen printed via pads, to allow all the wiring to originate from a central point.

8.3.5 Recorded signals from Frank configuration electrode network

The signals recorded using the printed Frank configuration garment are compared to standard commercial electrode Frank configuration recordings available online. The PTB database on physionet.org [79] contains 12-lead and Frank lead recordings for 300 patients, around 50 of whom were healthy patients, included in the database as control subjects. Of these, patient 156 was selected as he was a young adult male, included as healthy control. It is therefore likely that the electrical behaviour of his heart will show some similarity to that of the author. Although no two ECG’s are exactly alike it is hoped that there will be some similarity between this online recording and data recorded with the Frank configuration garment. The Frank configuration recording for patient 156 is shown in Figure 8.3-11. The voltage $V_Y$ has been flipped vertically to ease comparison with the recorded data. This is effectively the same as swapping the differential electrodes on a bipolar recording and does not affect the observed results in any other way.
Figure 8.3-11: Frank configuration recordings from a 1992 recording session with a healthy, 17 year old male, from Physionet.org [79].

For these recordings a resistive network was fabricated on strip-board using the resistive network design proposed by Frank, with the resistive constant $R$ set at 100 kΩ, based on the recommendations of the same paper [16]. The resistive network used 4 mm test connectors to connect to the amplifier cables and screw terminals to connect to wires, which are ultimately connected to the commercial electrodes or printed electrode vest. The fabricated resistive network and the resistive network schematic are shown in Figure 8.3-12 and Figure 8.3-13 respectively.

First, to ensure the resistive network was operating correctly, a Frank configuration measurement was taken from the author using 3M RedDot repositional Ag/AgCl electrodes. A Y axis measurement was taken both with electrodes on the foot and neck, as recommended by Frank, as well as 10 cm above and 20 cm below the heart, on the front of the chest, as used in the printed network. Three channels of the standard biopotential amplification apparatus...
were used for this measurement. The signals recorded using the positions proposed by Frank are shown in Figure 8.3-14 and those using the vest configuration from this thesis are shown in Figure 8.3-15. No digital filtering is used for the recorded signals in this chapter.

![Figure 8.3-14](image1)

Figure 8.3-14: X, Y and Z axis ECG recordings, taken using commercial Ag/AgCl electrodes in the positions recommended by Frank, and amplified with the standard biopotential amplification apparatus. No digital filtering is used.

![Figure 8.3-15](image2)

Figure 8.3-15: X, Y and Z axis ECG recordings, taken using commercial Ag/AgCl electrodes in the modified Frank positions described in the brief of the BRAVEHEALTH project, and the standard biopotential amplification apparatus. No digital filtering is used.

The two Frank configuration recordings with commercial electrodes are very similar. The main difference is in the V_y measurement, where the QRS peak has an amplitude of around 1 V.
the recording with Frank’s recommended electrode positioning, while the same deflection is greater than 1.5 V when the head and foot electrodes (H and F) are positioned on the chest, above and below the heart respectively. This is caused by a reduction in the impedance between the heart and the area of skin from which the measurement is taken. A larger $V_Y$ potential will be observed with the modified electrode configuration irrespective of the electrode type used.

A recording was then taken using the printed Frank configuration vest. The vest was worn for 1 minute before the recording was initiated. No conductive gel or paste was used and the signal is not digitally filtered. The X, Y and Z axis ECG potentials are shown in Figure 8.3-16.

![ECG recordings](image)

**Figure 8.3-16:** X, Y and Z axis ECG recordings, taken using the Frank configuration vest fabricated by screen and stencil printing, and the standard biopotential amplification apparatus. No digital filtering is used.

This recording shows the same deflections that are observed in the previous two measurements with commercial electrodes. The $V_Y$ QRS amplitude remains around 1.5 V, similar to that observed with Ag/AgCl electrodes in the modified Frank configuration. There is more noise in the modified configuration, especially on the $V_Z$ recording. This may be due to contact issues with electrode E, which is positioned over the chest cavity and has a greater contribution to $V_Z$ than to $V_X$ or $V_Y$, as shown previously in the resistive network schematic, Figure 8.3-13.

In terms of correlation with the recording of patient 156, there are several important similarities. First, the amplitudes of deflections for each axis are in similar proportions, with $V_Y$ being the largest deflection in all recordings shown here. However, $V_X$ and $V_Z$ have roughly the same amplitude in recordings taken in this project, while $V_Z$ is significantly larger in the recording of patient 156. The $V_Z$ printed vest recording also shows good agreement with that
of patient 156, especially the P and T deflections, before and after the QRS peaks, which are inverted relative to the QRS peaks in all recordings of V2 shown in this section.

8.3.6 Conclusion

This section has demonstrated the use of screen and stencil printing to fabricate a working Frank configuration monitoring garment. The garment allows the electrical potential from the heart to be monitored in three different orientations with minimal setup time and discomfort.

There are several problems with the garment structure. The garment could be made more comfortable by a designer with expertise in this area. The garment’s stretchability is low and consequently arm movements can cause motion artefacts in the recorded signals. The electrode size could also be increased to reduce electrode-skin impedance and thereby reduce noise.

In an application setting several different sizes of the garment would be required to fit various body types. The number of different sizes needed to serve the majority of the population would be lower if the garment had greater stretchability. Increasing the stretchability of the conductive tracks would allow one garment to fit various body sizes.

Noise due to electrode impedance mismatch is observable on the recorded signals. A settling time (<3 minutes) is also required to bring the baseline drift to manageable levels. Both issues could be solved using the printed active electrodes developed in this thesis.

8.4 Facial EMG/EOG headband

8.4.1 Introduction

The one-lead chest band and Frank configuration ECG electrode networks already described in this chapter have large (10-100 cm) electrode spacing. There are also several applications of skin surface electrodes that require a denser (1-10 cm) electrode spacing, for the analysis or stimulation of skeletal muscles (EMG), recording brain signals (EEG) or for a high density ECG. This and the next section both describe systems which aim to demonstrate the applicability of the technology developed in this thesis to high electrode density applications.

This section discusses a denser electrode network for facial EMG and EOG based on the technology described in this thesis. This was developed with the help of Fan Cao who worked on this project during his summer internship, however all the recordings shown here are from the author. Systems that use facial EMG and EOG signals for computer control [80] and as a wheelchair interface [81] have been reported. These systems typically use commercially available Ag/AgCl electrodes although Vehkaoja et al [82] report a 5 electrode system using embroidered silver coated fibres. A system that uses printed electrodes or dry rubber electrodes on textile for this purpose has not been previously reported.

Facial EMG is the monitoring of electrical potentials observable when muscles in the face contract. EOG is the monitoring of the corneo-retinal standing potential that exists between the front and back of the eyes. This electrode network is fabricated as a band worn around the head, with electrodes on the forehead. This minimises user discomfort, by removing the need
for a chin strap or electrodes and wires on other parts of the face used in some contemporary systems [83].

8.4.2 Goals of the system

Using the biopotentials observed on the forehead, the device aims to create an interface between a person and a computer using only facial muscles. Such systems are used in assisted living but printed textiles have not previously been used for this purpose. In this implementation, facial muscles are used to control the movement of a mouse cursor.

8.4.3 Layout of the facial EMG/EOG headband

Based on work by Law et al [84] and McMinn’s colour atlas of head and neck anatomy [85], it was discerned that the frontalis, which contracts to raise the eyebrows, the temporalis, which is one of the muscles used to clench the jaw, and the lateral rectus, which controls horizontal movement of the eye, are all well positioned for monitoring by electrodes on the forehead. Four electrodes are used in the designed system to monitor these muscles. The positions of the described muscles and the positions of the electrodes that are used to monitor them, are shown in Figure 8.4-1.

![Figure 8.4-1: The position of the muscles of interest and electrodes in the facial EMG system.](image)

Initially, the wearable electrode network was designed with ten pairs of electrodes. It was found that this number of electrodes resulted in significant motion artefacts which were synchronised on all electrodes. Unlike adhesive electrodes, which are usually mechanically independent from one another, the dry textile electrodes are mechanically connected via the textile garment. Movement under electrodes on the eyebrows caused the whole electrode network to move. Consequently motion on one electrode affects the other electrodes in the wearable network. The number of electrodes was reduced from twenty to four, which were placed on the forehead in positions where less movement was observed during facial movements. Figure 8.4-2 shows the position of the electrodes on the forehead. Electrodes that were removed from the wearable network to reduce motion artefacts are shaded.
The electrodes 2A and 2B are designed to be placed on the author’s temples when the headband is worn. Although the printed part of the device does not need to cover the full circumference of the head, it was necessary to place twenty printed electrodes on the forehead, as well as having twenty printed via connections to external electronics on the sides or back. This exceeds the maximum print width of 20 cm. It was necessary to use two separately printed structures to provide the required width. This provided a set of twenty electrode pads in the middle with ten via pads at either side. Electrode pads and via pads were both circles with a 10 mm diameter. The conductive layer layout for the twenty electrode headband is shown in Figure 8.4-3.
8.4.4 Fabrication of facial EMG/EOG headband

The screen design was printed on to two patches of Escalade. These were then cut into patches measuring 20 x 6 cm. Electrodes were stencil printed on to opposite sides on each of these. A printed Escalade patch after stencil printing electrodes is shown in Figure 8.4-5.

The two screen printed textile patches are sewn on to a 60 x 600 mm strip of Elasta along with hook and loop textile to secure the band around the head. Figure 8.4-6 shows the fabricated ten electrode headband.
If the screens were redesigned for a four electrode system it would be possible to have all the via connections in a single location, and to have the electrode network on a single screen. For a prototype system, modifying the original headband for use with fewer electrodes by changing the locations that are stencil printed is sufficient. Figure 8.4-7 shows the second revision of the headband, with only four electrodes.

8.4.5 Recorded signals from facial EMG/EOG headband

The signals recorded by the electrode pairs during each muscle contraction are given here. The standard biopotential amplification apparatus is used. There is no digital filtering on these signals and no conductive gel or paste is applied. The signal recorded during frontalis contraction (raising the eyebrows) as recorded on electrode pair 5A-5B is shown in Figure 8.4-8.
Figure 8.4-9 shows the signal recorded during temporalis contraction (jaw clenching) on electrode pair 2A-2B.

![Signal recorded on electrodes 2A-2B during a temporalis contraction (clenching the jaw). No digital filtering is used.]

Figure 8.4-9: Signal recorded on electrodes 2A-2B during a temporalis contraction (clenching the jaw). No digital filtering is used.

Figure 8.4-10 shows the signal recorded on electrode pair 2A-2B during left and right eyeball rotation.

![Signal recorded on electrode pair 2A-2B during left and right eyeball rotations (horizontal EOG). No digital filtering is used.]

Figure 8.4-10: Signal recorded on electrode pair 2A-2B during left and right eyeball rotations (horizontal EOG). No digital filtering is used.

It is clear the frontalis and temporalis signals are different in nature from the horizontal EOG (hEOG) signal. The frontalis and corrugator signals are AC signals, while the hEOG is a DC signal. This affects the way features are extracted.

### 8.4.6 Operation of facial EMG interface

The electrodes 2A, 2B, 5A and 5B in Figure 8.4-2 are arranged into two channels, 2A-2B and 5A-5B. Due to their positions, temporalis muscle contractions give larger signal amplitudes on 2A-2B than 5A-5B. The signal amplitude during frontalis contraction is around the same amplitude on both electrode pairs. Horizontal eye rotation is observable as a DC potential on electrode pair 2A-2B. Figure 8.4-11 shows how the signals are amplified and filtered to give three outputs, each designed to indicate the contraction of an individual muscle.
It was decided that the most intuitive method to control movement with these signals would be for the position of a muscle on a face to correspond with the direction of the cursor movement its contraction initiates. Consequently, a frontalis contraction (eyebrow raising) initiates upward cursor movement and a temporalis contraction (jaw clenching) initiates a downward cursor movement. The horizontal cursor movements are initiated by the corresponding eye rotations; looking left will start the cursor moving left and looking right will start the cursor moving right.

The hEOG signal, shown above in Figure 8.4-10, can be detected by a simple peak detector, however to improve accuracy when there is signal baseline drift the gradient of the signal is used. A sufficient change in signal amplitude during a given period of time will initiate a left or right movement. In a calibration trial, a horizontal signal was defined as a signal deviation of 175 mV after amplification within 200 ms.

Because the frontalis muscle covers most of the forehead, its contraction causes an AC signal on electrode pairs 2A-2B and 5A-5B. The temporalis, however, is on the side of the face and therefore causes a larger signal amplitude on 2A-2B. Consequently, both channels are monitored to identify a signal. The relative magnitudes of the AC signals on electrode pairs 2A-2B compared with 5A-5B show whether an AC signal is due to temporalis or frontalis contraction. The interface waits for the combined magnitude of \( V_{\text{jaw}} \) and \( V_{\text{eyebrow}} \) to exceed a threshold value \( (T_{\text{ON}}) \). When the threshold is exceeded, the ratio between the RMS values of \( V_{\text{jaw}} \) from pair 2A-2B, and \( V_{\text{eyebrow}} \) from pair 5A-5B, is calculated. Above a given ratio threshold \( (T_{\text{RATIO}}) \), the signal is defined as a frontalis motion, otherwise it is assumed to be a temporalis contraction. The threshold value \( T_{\text{ON}} \) is 0.12 V and the ratio threshold \( T_{\text{RATIO}} \) is 0.3. The rules for distinguishing between frontalis and temporalis contractions to provide up and down commands are summarised in Equation 4.
When $V_{\text{EYEBROW}} + V_{\text{JAW}} > T_{\text{ON}}$:

- Up: $V_{\text{EYEBROW}}/V_{\text{JAW}} > T_{\text{RATIO}}$
- Down: $V_{\text{EYEBROW}}/V_{\text{JAW}} < T_{\text{RATIO}}$

Equation 4: Rules for the detection of frontalis contraction (eyebrow raising - ‘up’ command) and temporalis contraction (jaw clenching - ‘down’ command).

A recording was taken using this wearable electrode network and the standard biopotential amplification apparatus without conductive gel or paste or digital filtering. During this recording the frontalis and temporalis muscles were both contracted twice. The RMS magnitude of the AC signal on the $V_{\text{EYEBROW}}$ (from 5A-5B) and $V_{\text{JAW}}$ (from 2A-2B) channels and the calculated ratios are shown in Figure 8.4-12.

**Figure 8.4-12:** Distinguishing recorded jaw clenching and eyebrow raising signals by the ratio of the RMS signal amplitudes on channel 2A-2B and 5A-5B. No digital filtering is used.

The four instructions are used to control movement of a mouse cursor around a monitor. If the cursor is still, a contraction will initiate movement in the intended direction. An instruction to move in the opposite direction to motion will stop the cursor. An instruction to move perpendicular to the direction of movement will cease movement in the initial direction. For example, if the cursor is moving up and a ‘left’ instruction is received, the cursor will cease to move up and begin to move left. The cursor speed is constant and defined in a GUI. The states used to control the cursor are shown in Figure 8.4-13.
8.4.7 Performance of facial EMG/EOG headband

EMG/EOG computer control is a relatively narrow field, and so there is no existing standard by which such systems can be compared. Here, the performance of the full system for mouse control is gauged by the ease with which a mouse can be moved to different points on the screen. A test image, which filled a monitor during testing, consisted of a grid of sixteen numbered squares. The test image is shown in Figure 8.4-14. A photo of the author testing the device is shown in Figure 8.4-15.

A sequence of ten numbers between one and sixteen was provided to the user. The length of time taken for the user to stop the cursor in the square corresponding to each number, in the correct order, was measured. The time taken was defined as the ‘task completion time’.

Five different cursor speeds were tried, with five trials of different number sequences at each speed. The same ten-number sequences were used at each cursor speed, so that there was no variation between total cursor travel distance at each speed. The average task completion times for each speed are shown in Figure 8.4-16.
It was found that 200 pixels/second was the optimal speed, although 150 and 300 did not have significantly increased task completion time. It was found that increasing the cursor speed made relatively long-distance movements, for example moving between grid square 13 and 8, faster. However, short distance movements, for example between grid square 13 and grid square 10, were slower and more difficult. User control of the cursor speed would improve the ease with which cursor position can be controlled.

8.4.8 Conclusions

This electrode network provides a wearable, reliable and reusable way to interface with a computer using only facial muscles. This is the first example of a printed textile used to monitor biopotentials originating in facial muscles. It also shows that the technology developed in this thesis is applicable to denser electrode networks.

Motion artefacts were encountered during the use of this device, especially during eyebrow raising. This occasionally resulted in instability on the hEOG channel which could cause false hEOG motion detection immediately after the eyebrow raising. This could be reduced with the introduction of a delay between an eyebrow raising instruction and the acceptance of any other instruction, but naturally this also reduces ease of use. No work has been done on calculating the false detection rate and the task completion times are inclusive of any false detection events.

There was some discomfort encountered during the use of this device, especially with the electrodes over the temples. By increasing the electrode size both the signal quality and comfort would be improved. Although the smaller electrodes were required for the high density of the initial 20 electrode network, significantly larger electrodes could be used for a four electrode network like that eventually adopted. A carbon loaded rubber with a lower Young’s modulus would also decrease discomfort.

The vias on this design are as large as the electrodes and consequently take up a large amount of screen space and materials. If networks are created with a larger number of electrodes a specially designed bus connection method may be more economical and convenient than the
individual electrodes. The initial design had high electrode density which meant the screen printed design covered a 20 cm width. This meant that the connection points had to be separated into two sites, which is not convenient for connection. An illustration of a headband with larger electrodes and connection points all in a single site is shown in Figure 8.4-17.

![Illustration of a headband with larger electrodes and connection points](image)

**Figure 8.4-17**: Revised layout for EMG/EOG headband, with larger electrodes and all vias on one side of the head.

The 50 Hz noise was low enough for this application even with small electrodes and unshielded printed connections. Because the signals are used for computer control rather than recorded for diagnosis, a bandstop filter around 50 Hz could be used to further reduce the noise. Generally the AC signals, from the frontalis and temporalis, could be detected more reliably than DC signals from the lateral rectus as they were not susceptible to false positives during high motion artefact. The implementation of active electrodes could improve this.

### 8.5 Bicep EMG armband

#### 8.5.1 Introduction

This section discusses the development of an arm band for EMG research. The electrode network is the largest developed here in terms of number of electrodes. Although the network comprises 16 electrodes in total, which could be arranged into a large number of amplified pairs, a maximum of three pairs is used here because the only amplifiers available were those that were fabricated specifically for this project. This was developed with the help of Dianzhong Chen, who performed the experiments described in this section on himself.

#### 8.5.2 Goals of the system

The aim of this system is to demonstrate the applicability of these electrode networks as a research tool and a method for prosthetic control. In this application, EMG signals originating in the upper arm are examined with an electrode network covering 10 x 10 cm. It is thought that these systems would facilitate easy research in the examination of the electrical activity associated with various movements because the electrode placement is predetermined and no skin preparation is required.

Previous research has provided some solutions to covering an area of the skin with a correctly-spaced high-density network of electrodes. Guzman *et al* [86] used a network set in plastic
with traditional Ag/AgCl electrodes that had to be chlorided after some use. This device will save on application time, but more specialist equipment is required to periodically re-apply the silver chloride layer by electroplating than is required for washing. Consequently each device of this type would require this re-chloriding equipment for its use. Guzman et al’s device is shown on-application in Figure 8.5-1.

![Arm band fabricated by Guzman et al](image)

**Figure 8.5-1: Arm band fabricated by Guzman et al [86].**

As in research by Guzman et al, this research will examine a particular phenomenon known as the innervation zone. The innervation zone, also known as the endplate zone, is the point on the muscle where the motor neuron joins. The initiation of the contraction of muscle fibres is delayed further from the innervation zone because the muscle fibres conduct with a fixed muscle fibre conduction velocity [87].

Consequently, the EMG signals measured from a pair of electrodes separated vertically on the upper arm will be reversed on either side of the innervation zone of the bicep, and the signal will appear to be delayed as the distance from the innervation zone increases. Easy detection of the innervation zone is especially useful where drug therapies are used to overcome spasticity in victims of neurological legions. The closer the drugs are injected to the innervation zone, the better recovery appears to be. As this position is different in different people, a way to easily detect the innervation zone is helpful. The innervation zone is marked as ‘ZI’ on Figure 8.5-1 and corresponds with channel 5 in Figure 8.5-2.
Signals recorded by Guzman et al. from a series of vertically separated electrodes on the upper arm. The innervation zone is located on channel 5 [86].

Showing the ability to locate the innervation zone demonstrates that these networks could be used as a research tool. Subsequently, the device is used to examine the effect of inter-electrode spacing and the strain of the motion performed on the observed signal.

8.5.3 Layout of EMG armband

This network is composed of 16 passive electrodes and conductive tracks connecting the electrodes to a centralised connection point. Vias are fabricated as previously described at this centralised connection point to permit the connection of shielded cables. The exposed conductive area of the via connections is 9 mm in diameter and the exposed conductive area on the electrodes is 11 mm in diameter. The carbon loaded rubber electrodes printed atop the screen printed electrode pads are 13.5 mm in diameter, 5 mm thick and have an inter-electrode spacing of 12 mm. The electrode centres are 25.5 mm apart. The screen designs AB-1 are shown in Figure 8.5-3 (a) to (c).
Figure 8.5-3: Screens for printing an EMG armband for monitoring the electrical activity of the bicep (AB-1). (a) is the interface layer, (b) is the conductive layer and (c) is the encapsulation layer.

8.5.4 Fabrication of EMG armband

The screen design AB-1 described above is used to print on to Lagonda textile. For convenience UoS-IF-#4 is used as the paste for the interface and encapsulation and Dupont 5000 is used for the conductive layer. The screen printed textile is shown in Figure 8.5-4.

Figure 8.5-4: Screen printed conductive network on Lagonda textile.
Carbon loaded rubber is then stencil printed on to the electrodes with a 13.5 mm diameter. The printed Lagonda textile is then sewn to the elastic textile to form a wide elastic band that can be wrapped around the arm. Silicone foam is placed between the two textiles to reduce the severity of pressure changes when the arm changes diameter. Hook and loop textile is attached to either end of this band so that it can be secured around the arm. Finally, the via pads are penetrated with steel buttons so that the electrodes can be electrically connected from the outside of the textile. The fabricated device is shown in Figure 8.5-5.

Figure 8.5-5: The completed arm band device.

8.5.5 Recorded signals from EMG armband

This section discusses the use of the fabricated arm band device to examine four factors affecting the EMG signal. These are the location of the innervation zone on the forearm, the effect of electrode orientation, the effect of electrode spacing and the applied force.

8.5.5.1 Experimental procedure

For each measurement described here, a specific motion was performed, so that the measurements are comparable. The arm was extended with the patch applied to the arm. The positioning of the patch will be discussed later as it is different for each 15 second measurement. The recording was started with the arm relaxed. After 4 seconds a light flashes to indicate that the arm should be lifted to a specified height. The time taken to lift the arm to this height is not controlled. The arm is then held at this height for the remainder of the 15 second measurement. An Ag/AgCl DRL electrode is placed on the forearm for all experiments in this section. The standard biopotential recording equipment is used. The experimental setup is shown in Figure 8.5-6.
Figure 8.5-6: The experimental setup for measurements with the electrode network armband.

For the location of the innervation zone, inspection of the raw signals from electrode pairs is sufficient. The other measurements, however, require examination of the average magnitude of the AC signal and so an RMS measurement is used. For all signals recorded with this device, a 2nd order Butterworth notch filter from 40 to 60 Hz is applied.

8.5.5.2 Innervation zone

The first experiment carried out is the location of the innervation zone, because this will guide the electrode placement in future experiments. The expected signal, based on channels 3, 5 and 7 in work by Guzman et al [86], previously shown in full in Figure 8.5-2, is shown in Figure 8.5-7. In this work the distance between the centres of electrodes is 10 mm.

Figure 8.5-7: Three channels around the innervation zone in work by Guzman et al. The innervation zone is located within channel 5 [86].

The innervation zone is located in this work by examining the signals from four electrodes arranged into three proximodistally separated pairs, as shown in Figure 8.5-8. The arm band is applied at different positions on the arm, with the sensing electrodes positioned on the anterior of the arm. When the innervation zone is located inside the area covered by electrode pair 2-3, the AC signal from electrode pair 1-2 will be an inverted copy of electrode pair 3-4. Electrode pair 2-3 will also have a lower RMS AC amplitude.
Figure 8.5-8: Electrode, amplifier and filter configuration for the detection of the innervation zone.

After moving the arm band network incrementally from the shoulder in a distal direction the armband electrodes were eventually positioned over the innervation zone and the signals shown in Figure 8.5-9 were obtained. A 50 Hz notch filter is applied. No conductive gel or paste is applied for any of the arm-band experiments described here.

Figure 8.5-9: Three signals recorded simultaneously with the textile electrode armband. Electrodes are aligned horizontally and separated proximodistally along the upper arm. The signal is passed through a 50 Hz notch filter.

These results show clearly that the signals from electrode pairs 1-2 and 3-4 are mirrored. The signal from electrode pair 2-3 has a relatively low amplitude and is a slightly out of phase version of electrode pair 1-2. This suggests that the innervation zone is located near electrode 3. It is clearly possible to detect the innervation zone with a system using these electrodes, although the detection precision would be improved with a denser electrode network, using both smaller electrodes and smaller inter-electrode spacing. Automatic detection could be integrated by detecting the channel with the lowest RMS signal amplitude over a period of time.
8.5.5.3 Effect of inter-electrode spacing on signal amplitude

This experiment examined the effect of increasing inter-electrode distance. As the innervation zone had been located, the arm band was situated on the proximal part of the upper arm to avoid the innervation effect. A simultaneous recording was taken with electrodes having inter-electrode spacing of 25 and 50 mm, which correspond to electrode pairs 1-2 and 1-3 as shown in Figure 8.5-10. 75 mm spacing was not examined because the size of the device caused electrode 4 to be positioned over the previously located innervation zone, which could affect the results.

Figure 8.5-10: Electrode, amplifier and filter configuration for the examination of the effect of inter-electrode distance.

Five recordings were made using the 15 second test regime described earlier and the RMS voltage on each channel was extracted from this data. The averaged results are shown in Figure 8.5-11. A 50 Hz digital notch filter is applied.

Figure 8.5-11: Average RMS voltage for printed textile electrodes situated on the proximal portion of the upper arm with inter-electrode spacing of 25 and 50 mm. A 50 Hz digital notch filter is applied.

The initial peak corresponds with the lifting of the weight up to the pre-defined height. The RMS voltage then drops to a lower level, as the bicep remains contracted to support the
weight. Both the peaks and the steady-state amplitudes are significantly higher than the RMS noise at rest.

It is clear from this result that the amplitude of a signal recorded with a larger electrode spacing is greater. However, the larger amplitude may not be associated with an increase in accuracy of detection as it is clear from this result that the RMS amplitude of the signal recorded with a 25 mm spacing is more stable, while the 50 mm spacing recording is more subject to changes in RMS amplitude during the 6-15 s period of the recording.

8.5.5.4 Effect of applied force on signal amplitude

The final experiment carried out with this system is an examination of the relationship between the force applied at the hand and the RMS amplitude of the EMG signal recorded from the textile arm band positioned in the same place as in the previous experiment. This experiment was carried out with the textile electrodes and a proximodistal inter-electrode distance of 25 mm. The electrodes were placed as shown in Figure 8.5-12.

The RMS EMG signal with 2 kg, 5 kg and 7 kg weights lifted are shown in Figure 8.5-13. A 50 Hz notch filter is applied.
There is a clear correlation between the weight lifted and the magnitude of the EMG signal. The stability of the RMS voltages recorded decreases with increasing weight.

8.5.6 Conclusions

This prototype demonstrates that the printed electrodes provide suitable signal output to be combined with skeletal muscle EMG acquisition technology and could be used in a more detailed, clinical research study. They would be especially applicable to ambulatory monitoring. Conclusions reached using the electrode network described here agree with those in the literature [86], suggesting that the signal quality with passive electrodes will be sufficient for some research tasks, especially those where DC changes causing baseline drift can be controlled by the amplifier design or reduced in filtering. For tasks that require greater signal stability, such as observing DC changes between paired muscles, active electrodes could be employed.

Prefabricated electrode networks have been described previously in the literature [86], [57] and are usually used to save time with studies using high density electrode networks. Even with the relatively small number of electrodes used in these experiments, the process of applying electrodes with a wearable band is significantly faster than individually placing self-adhesive electrodes, reducing the time taken for a given procedure. If multiple subjects are used there is little chance of variability in electrode placement between subjects as the electrodes are permanently positioned relative to one another, improving the reliability and reproducibility of the data produced and reducing the possibility of human error. The comfortable and dry non-degrading nature of the electrode networks mean that they can be used for long-term, ambulatory applications. The device described in this section differs from previous devices in that the electrodes do not need to be replaced between each use, which will also save time and expense.

8.6 Functional electrical stimulation system based on this work

Electrodes fabricated using the technology developed in this thesis have been employed in the University of Southampton’s Institute of Sound and Vibration Research (ISVR) for functional electrical stimulation (FES). This involves passing current through the arm via two electrodes and observing with a high frame rate camera how muscles contract. This information can be helpful in rehabilitation after loss of motor function. The electrode network, designed by Dr Kai Yang and Dr Russel Torah, is based on the research in this thesis. It is shown in Figure 8.6-1. This system is described in more detail in a published journal paper [87].
8.7 Conclusions

Each of the electrode networks described in this chapter have provided insight into how these systems might be applied in diagnosis, clinical trials, human-computer interfaces, rehabilitation, and ambulatory or remote patient monitoring.

The active electrode chest band demonstrates the integration of passive and active electronic components into a deposited electrode structure. It also shows that an ECG signal comparable in quality to that obtained with Ag/AgCl electrodes can be acquired when active electrodes are fabricated using the deposition processes described in this thesis. The novel structure of the active electrodes, which contains all components within the electrode, is implemented with a homogeneous fabrication technique. The fact that the components are inside the relatively rigid electrode rather than exposed on a separate portion of the textile could, with the modifications discussed, offer high durability. There is no technical barrier that prevents these electrodes being implemented in any of the other systems described here; they would have been implemented if the active electrodes had been developed earlier in this work.

The Frank configuration vest demonstrates the suitability of screen printing for fabricating large area wearable electrode networks. This is the first Frank configuration vest garment fabricated with printing or any other smart textile technique as far as the author is aware. It provides signals with all expected peaks visible on all three channels. Thin layers of silver can be deposited over a large area in very little time, whereas other printing methods that use a movable nozzle, such as inkjet or dispenser printing, would take significantly longer. The combination of inelastic and elastic textiles, as well as silicone foam beneath the electrodes, meant that X, Y and Z Frank configuration leads could be recorded from the author despite the fact that dry, passive electrodes were used. The setup time for this garment is less than individually taping electrodes over the body, however there is a settling time (<3 minutes) before baseline drift falls to acceptable levels. Using active electrodes would remove this settling time and reduce the level of 50 Hz noise and baseline drift observed when this
wearable electrode network is used, thus significantly reducing setup time and improving signal quality and reliability for clinicians and researchers.

The facial EOG/EMG headband demonstrated the suitability of this printed textile technology for EOG and EMG feature extraction. The passive electrodes used had no issues with observing the AC EMG signals from the frontalis and temporalis muscles, but there were occasionally false positives in the feature extraction of the horizontal EOG, a DC signal. These were caused by motion artefacts when the eyebrows were raised. The performance of this system could be improved by reducing the false detection rate. Implementing active electrodes could improve performance, but increasing the size of the electrodes as shown in Figure 8.4-17 or placing silicone foam beneath them as shown in Figure 8.3-8 could offer sufficient reductions in motion artefact.

The bicep EMG armband demonstrated that printed textile electrode networks have the potential to be used in research applications. However, due to the large dimensional change at the bicep during contractions, there were significant pressure changes at the electrodes and consequently there were severe motion artefacts on the recorded signal, similar to those shown in section 7.3.5. These are not shown in the graphs shown here because these graphs show only the RMS noise and do not show DC changes. The pressure changes were not reduced sufficiently by the inclusion of silicone foam beneath the electrodes.

The key result from this demonstrator was that the level of dimensional change at the observed body part is a large factor in determining the baseline stability of a signal recorded with an elasticated monitoring system. Greater garment stretchability should be a feature of monitoring garments recording biopotentials at sites where there is large dimensional change. This would reduce pressure changes at the electrodes and thereby reduce motion artefact, but would require conductive tracks with greater stretchability, such as those described on page 49 in section 3.4.2.

As already discussed, the active electrodes have clear advantages over the passive electrodes, although for some applications the passive electrodes perform sufficiently well and the use of active electrodes would be unnecessarily complicated and expensive. In the feature extraction of frontalis and temporalis contractions described in section 8.4, the passive electrodes perform sufficiently well. For diagnostic applications, where signal filtering must be minimised, active electrodes are clearly preferable.

The electrode network methodology in this thesis has been used by other researchers in the department both as monitoring electrodes and for functional electrical stimulation. The facial EMG and EOG system and the bicep EMG system were developed as part of internship and MSc projects respectively.
9 – Conclusions and Future Work

This chapter concludes the thesis by summarising the work carried out, discussing its significance in the wider context of wearable biopotential monitoring and recommending directions for future research in this area.

9.1 Literature review

The literature review in this thesis examined the limitations of existing wearable biopotential monitoring systems. Most of these systems were implemented with conductive yarn electrodes or standard Ag/AgCl electrodes. Conductive yarn electrodes are generally limited in terms of their performance because they have high electrical impedance and offer no control over the DC potential at the electrode, resulting in high baseline drift. Ag/AgCl electrode systems perform well in terms of signal acquisition, but Ag/AgCl electrodes gather hair and dirt to their adhesive layer and have to be replaced daily. Printed wearable biopotential monitoring systems have not been investigated to the same extent and have advantages over Ag/AgCl electrodes in terms of compatibility with textiles and have advantages over conductive yarns in terms of ease of manufacturing and compatibility with planar electronics. Although there are examples of printed passive and active electrodes in the literature, these were implemented using conductive yarns to connect the electrodes. Wearable biopotential monitoring electrode networks with predefined electrode positions in which the electrodes and conductive paths are both printed have not been previously investigated.

Contemporaries working towards the aim of printed wearable biopotential monitoring focus on printing their systems onto non-woven textiles. Woven textiles are preferable for long term wearable monitoring as they are breathable and comfortable to wear, whereas non-woven textiles such as Tyvek and Evolon have a texture more like paper and are typically far less breathable. A polymer interface layer on a woven textile can be used to create a smooth surface for printing subsequent functional layers. The literature review also showed that an encapsulation printed or laminated on top of the functional layer or layers greatly improves durability.

9.2 Process selection

Screen printing is a suitable technology for the fabrication of textile based electrode networks for biopotential monitoring. It is the most appropriate deposition method for quickly printing large areas (>100 cm²) with thin layers (<10 μm), as are required for wearable electrode networks. This advantage is more obvious with a larger area, for example in the fabrication of the Frank configuration vest. Inkjet printing could save on cost by reducing the thickness of layers but the flow rates are typically lower than screen printing and consequently the manufacturing process would be time consuming. It is also much harder to achieve consistent prints with homogenous layer thicknesses using inkjet printing on textile.

Screen printing also has several advantages compared with conductive yarn, which is the most popular technology for the fabrication of textile based electrodes and connections. The first
advantage is the same as that described above; screen printing is a faster fabrication technique. Conductive yarns also tend to have higher resistivity than printed conductive tracks because they have many constraints on their properties. For comparable performance with deposition techniques, conductive yarns should be insulated. They also need far greater flexibility than printed tracks as they need to be woven or embroidered by high speed sewing machines. Insulated conductive yarns are more difficult to connect to planar electronics and usually have to be stripped and connected manually, whereas screen printing an interface layer allows planar electronic circuits to be integrated with connections to electrodes in a more homogeneous fabrication process. Compatibility with planar electronics is, then, another area in which the performance of printed textiles surpasses that of conductive yarns. Screen printing was deemed the most appropriate technology for textile based wearable biopotential systems.

9.3 Material selection

Screen and stencil printed pastes were examined to implement the designed structures. Four screen printable polyurethane pastes developed by the University of Southampton were compared through a series of tests in section 4.2. It was found that UoS-IF-#35 and UoS-IF-#44 had worse mechanical properties than UoS-IF-#4 and UoS-IF-#39. The latter two pastes were selected for further examination for the fabrication of conductive track structures durable to machine washing.

Four screen printable silver polymer pastes were compared to find the most durable in section 4.3. Electrodag 725A was found to have the lowest resistivity, at 0.056 Ω/square, and the highest durability, with a 358% resistance increase after 200 cycles internal bending. All three commercial pastes that were examined had suitable properties to be used in a wearable biopotential monitoring network with resistance increase of >700% resistance increase after 200 cycles. Dupont 5000 was used as this paste was more easily available, and its moderately increased resistivity compared to Electrodag 725A, 0.086 Ω/square, would not affect the performance of biopotential monitoring systems. Using this paste also allowed changes in durability to be gauged more easily in later experiments.

Four different silicone rubbers were examined in section 4.4 for potential use in a carbon loaded rubber for stencil printing at electrodes. Viscolo 22 was found to offer the best combination of tensile properties with a Young’s modulus of 0.25 MPa, the lowest of those tested, stencil printable viscosity and a conductivity of approximately 50 Ω/square. Introducing carbon increased the viscosity of the mixture, so two thinners were compared for use in formulating a lower viscosity carbon loaded rubber paste. T-402 solvent thinner was selected as it improved the resistivity, with an addition of 10 phr causing a 50% drop in the resistance of carbon loaded RTV-M silicone, and had no noticeable detrimental effect on tensile strength in small proportions, with RTV-M silicone having roughly the same tensile strength with a 10 phr T-402 addition.
9.4 Durability

The one area where conductive yarns are superior to printed conductive textiles is durability. Consequently, the durability of printed systems has been the subject of investigations in this thesis and in recent literature.

Work was carried out on improving the durability of printed conductive track structures in chapter 6. After an optimisation process involving mandrel testing and machine washing, a significant improvement in durability was achieved from a resistance increase of 1000% after 100 mandrel cycles in initial tests to a 20-30% increase after 100 mandrel cycles for the final design, structure 5.b described in section 6.3.2. In initial washing machine tests 94% of encapsulated conductive tracks were broken after 10 machine washes. Using a structure composed of both UoS-IF-#4 and UoS-IF-#39, conductive tracks were printed with a width of 3 mm and a printed thickness of 320 µm. These were put through 10 machine washes and only one sample (out of 35) became non-conductive during this test. This one sample showed no rupture along its length and it is thought that the reason for its failure was the method of encapsulation of the vias. The average resistance increase of the printed conductive tracks that remained conductive was 192%. This would be acceptable for most biopotential monitoring applications and could be reduced by using a different silver polymer paste.

Mandrel testing showed that bending the conductive track normal to the plane had a different effect depending on whether the track was bent internally or externally. The neutral axis is in a different position in these two bending modes so there is no clear optimal position for the conductive layer. Consequently it was suggested that the interface/encapsulation structure with a combination of materials would be advantageous, as the UoS-IF-#39 with a lower Young’s modulus deforms to prevent excessive deformation of the conductive layer, which is enclosed in a beam of UoS-IF-#4 which has a significantly higher Young’s modulus. The stiffer beam moves towards the neutral axis, which is in different positions in external and internal bending. It was shown in section 6.3.1 that a design composed of a 1:1 ratio of UoS-IF-#4 and UoS-IF-#39 had better durability than those composed of a 1:0, 3:1, 1:3 or 0:1 ratio. This result is useful for the future design of polymer-encapsulated functional tracks on textiles to minimise damage to functional materials.

Increasing the flexibility of conductive tracks and the adhesion of electrodes is necessary to improve durability to the point where these electrode networks could be deployed for healthcare applications. Methods to improve durability with material development and design changes are discussed in the future work (section 9.8).

9.5 Electrode design

Designs were proposed in chapter 3 for conductive tracks, electrodes and garments for wearable biopotential monitoring networks. The textile printed conductive tracks were composed of a thin, 5 µm conductive layer on a polymer interface layer, enclosed by a polymer encapsulation layer. The passive electrode design used a stencil printed conductor loaded rubber to increase the thickness of the electrodes and improve contact stability between the electrode and skin. The novel active electrode design enclosed the required buffer amplifier
within the electrode and required printing on only one side of the textile. The garment was composed of a printed inelastic textile, which prevented excessive deformation at the printed structure, sewn to an elastic textile which allowed wearable garments and their electrodes to adhere to the skin without discomfort or adhesives. Textiles were compared and selected for this purpose. The careful design of versatile garments which provide surfaces to integrate screen printed electrode networks, stable electrode contact as well as comfort and breathability allowed later prototypes such as the Frank configuration vest to be fabricated.

9.6 Electrode performance

Ambulatory ECG monitoring is the main target application for the electrode networks described in this thesis. Existing systems for ambulatory ECG monitoring use adhesive Ag/AgCl electrodes that are attached to the chest and connected with standard wiring to an electronics box. The setup is expensive, because the electrodes must be periodically replaced, and uncomfortable, especially during sleeping. Further, the recorded biopotentials are not reliable for extracting morphological information because the electrode placement is defined by the user.

The performance of passive and active electrodes was examined in chapter 7. The passive electrodes had relatively low 50 Hz noise when optimised, 5 mV at a gain of 1000 with a 7 cm² contact area, but were susceptible to DC drift, especially during pressure changes at the electrode. They are consequently suitable for the recording of AC biopotentials, where DC drift can be filtered out. The susceptibility to DC drift also makes the passive electrodes more difficult to use in DC EMG applications, such as EOG, but they can be used in this scenario with appropriate filtering and feature extraction methods. For diagnostic ECG applications these passive electrodes cannot be used, as heavy filtering reduces the diagnostic usefulness of the signal.

Active electrodes, examined in section 7.4, showed very low levels of 50 Hz noise, 3 mV at a gain of 1000 with a 5.7 cm² contact area compared to 2 mV for commercial Ag/AgCl electrodes of approximately the same dimensions. 50 Hz noise was not even visible on recordings taken with these electrodes, meaning it is lower than the ADC resolution of 10 mV after amplification. Given that QRS peaks had an amplitude of around 1 V this equates to a signal to noise ratio of around 100:1. Motion artefacts were significant on these electrodes but it is thought that this may be more due to the mounting of the electrodes than the electrodes themselves. The active electrodes also offered greatly improved resistance to DC drift. The active electrode standard deviation was 0.04 V compared to 0.2 V for passive electrodes with the same contact area. Improved garment design to increase contact stability would yield wearable printed textile active electrode networks with performance equivalent to current clinical systems that use disposable Ag/AgCl electrodes.

The active electrode systems proposed in this thesis offer the potential to improve comfort without reducing signal quality. Since only a small area of the textile is printed the printed electrode network is breathable and comfortable. Fixed electrode distances also allow morphological data to be extracted with greater reliability than electrodes placed by an unskilled user.
The electrode performance evaluation in this thesis was not clinically acceptable, as all trials were carried out on the author. Variations in skin impedance between different users would affect the performance, at least of the printed passive if not the printed active electrodes. More rigorous trials using a group of subjects would make the results more reliable as a gauge of electrode performance. However, the performance evaluation here is sufficient to show that these electrodes have great potential for use in wearable monitoring systems even where high signal quality is required.

9.7 Prototype systems

The technology developed in this thesis was demonstrated in chapter 8 with four separate wearable biopotential monitoring systems using printed electrodes and conductive tracks. The first system used a wearable network of two active electrodes and a passive electrode on a chest band to monitor a one-lead bipolar ECG and had a very low level of 50 Hz noise. The second system used a wearable network of seven passive electrodes on a vest to successfully monitor a Frank configuration vectorcardiogram more easily than with standard electrodes. The third system used four passive electrodes on a headband to monitor muscles in the forehead to successfully control a computer mouse cursor. The fourth system used a grid of passive electrodes on an armband to examine the electrical behaviour of the bicep, with results agreeing closely with systems reported in the literature using Ag/AgCl electrodes.

The mix of biopotential monitoring demonstrators developed in this thesis show that the technology is versatile and applicable both to large separation electrode systems and high density electrode systems. They give high signal quality, require no electrode paste and are comfortable. They are consequently appropriate for a large number of long and short term ambulatory applications, from heart monitoring to prosthetic control to assisted living. This work has gone beyond the pre-existing state of the art and demonstrated the feasibility of a new technology for ambulatory biopotential monitoring. This work has resulted in 3 peer reviewed journal publications and 2 conference publications, with 2 further papers written and awaiting publication.

9.8 Recommendations for future work

The novelties and publications arising in this thesis will have a positive impact on the development of wearable electrode networks for ambulatory applications. This final section discusses recommendations for future work to improve these prototype electrode networks for improved durability and electrode performance.

Materials for use in printed textile electrode networks should be further explored and developed. This applies to both printable pastes and textiles. A polyurethane interface paste with a high Young’s modulus, like UoS-IF-#4, but with linear-elastic tensile behaviour, like UoS-IF-#39, would reduce the severity of ruptures and resistance increases in printed composite conductive tracks during bending normal to the textile. An exhaustive evaluation of silver polymer pastes would be necessary to ensure maximum durability but ultimately the durability of the printed conductive tracks is dependent on the mechanical properties of the interface and encapsulation layers. Because these silver polymer pastes are currently for the printed
electronics market and are not aimed at flexible fabrics, a bespoke paste developed specifically to be as flexible and durable as possible would seem the ideal solution.

The carbon loaded rubber paste developed in this thesis is optimised for the materials tested, but there were shortcomings in this optimisation process in that the initial selection of silicones was fairly limited. Since Viscolo 22 has been selected as the silicone rubber, other silicones with similar tensile properties but lower viscosity should be sourced and tested. There was also only one type of carbon black used in the optimisation. Further optimisation of this paste would require testing of a large number of formulations of carbon black and silicone rubber which could lead to an improved electrode conductivity and potentially provide a reduced Young’s modulus and therefore a more comfortable and flexible electrode.

The printed textile electrode networks in this thesis are printed on to an inelastic textile, so the electrodes are mechanically connected. Motion on one electrode can cause artefacts on other electrodes in this kind of mechanically connected system. Further, the distance between electrodes does not change and so the designs of wearable electrode networks have to be altered to fit different body shapes. Stretchable connections using printing on to woven textiles need to be investigated, as these would both reduce motion artefact and increase the range of people whom the garment would fit.

The durability of electrodes must be further improved so that printed textile electrode networks can be machine washed. The designs described in appendix D provide a starting point for further development. An improved electrode design may require the introduction of different printing techniques, such as dispenser printing or injection moulding. The durability of conductive tracks could also be further improved by reducing the thickness of the interface layer and moving the conductive layer closer to the surface of the textile. Mandrel testing at lower radii of 1 and 2 mm as well as 3 mm would provide a better understanding of the behaviour of conductive tracks during bending normal to the plane. A finite element model of these structures, drawing from the practical results described in chapter 6 would allow the behaviour of different structures to be predicted and negate the need for the rigorous optimisation in future applications with different materials.

The active electrode and amplifier circuits used in this thesis are based on work by other researchers and have not been optimised for this technology. An investigation into the optimal active electrode and amplifier circuits for different biopotential sensing applications would make this technology more useful for each application. The integration of electronics into printed textile electrode networks would be preferable, in order to make monitoring as portable and unobtrusive as possible, but would add a further challenge to durability. This would negate the need for textile vias.
References


thin chip package (UTCP) and stretchable circuit technologies for wearable ECG system,” in *2011 Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, Boston, 2011.


Appendix A – Material Properties

The properties of the textiles, interface pastes and conductive pastes used in this thesis are included here for reference. The properties of the formulated carbon pastes are not included.

**Textiles:**

<table>
<thead>
<tr>
<th>Textile</th>
<th>Escalade</th>
<th>Lagonda</th>
<th>Elasta</th>
<th>Elastic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Supplier</td>
<td>Klopman</td>
<td>Klopman</td>
<td>Elasta</td>
<td>FabricLand</td>
</tr>
<tr>
<td>Thickness (µm)</td>
<td>410</td>
<td>290</td>
<td>890</td>
<td>1035</td>
</tr>
<tr>
<td>Resistance to Temperature</td>
<td>Fine up to 130 ºC</td>
<td>Fine up to 130 ºC</td>
<td>Warping around 90 ºC</td>
<td>Warping around 90 ºC</td>
</tr>
<tr>
<td>Extension at break (%)</td>
<td>Warp: 48</td>
<td>Weft: 10.5</td>
<td>Warp: 13.5</td>
<td>Weft: 49</td>
</tr>
<tr>
<td>Young’s Modulus (N/cm)*</td>
<td>3.14</td>
<td>13.74</td>
<td>1.93</td>
<td>10.27</td>
</tr>
</tbody>
</table>

*For inelastic textiles the Young’s modulus at break is given. For the elastic textiles, the Young’s modulus is taken from the elastic (straight line) portion of the tensile characteristic at 100% extension.

**Interface pastes:**

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Curing time (s)*</td>
<td>20</td>
<td>60</td>
<td>40</td>
<td>20</td>
</tr>
<tr>
<td>Viscosity (Pa.s)</td>
<td>6.2</td>
<td>3.6</td>
<td>11.1</td>
<td>15.5</td>
</tr>
<tr>
<td>Deposits required</td>
<td>13</td>
<td>18</td>
<td>13</td>
<td>13</td>
</tr>
<tr>
<td>Contact angle</td>
<td>44.8</td>
<td>94.4</td>
<td>102.4</td>
<td>100.5</td>
</tr>
<tr>
<td>Scratch test (% increase)</td>
<td>16.3</td>
<td>13.2</td>
<td>123.2</td>
<td>36.6</td>
</tr>
<tr>
<td>Young’s Modulus</td>
<td>0.228</td>
<td>0.072</td>
<td>0.015</td>
<td>0.156</td>
</tr>
<tr>
<td>Elongation at break (%)</td>
<td>33.24</td>
<td>13.54</td>
<td>84.80</td>
<td>25.28</td>
</tr>
<tr>
<td>Water absorption (%/day)</td>
<td>21.1</td>
<td>0</td>
<td>0.7</td>
<td>0</td>
</tr>
</tbody>
</table>

*at curing power density 31mW/cm² at 365nm.

**Conductive pastes:**

<table>
<thead>
<tr>
<th>Paste</th>
<th>Dupont 5000</th>
<th>Electrodag 725A</th>
<th>Conductive Compounds Ag-800</th>
<th>UoS-TC-32</th>
</tr>
</thead>
<tbody>
<tr>
<td>Curing temperature (ºC)</td>
<td>120</td>
<td>120</td>
<td>130</td>
<td>120</td>
</tr>
<tr>
<td>Viscosity (Pa.s)</td>
<td>3.742</td>
<td>5.419</td>
<td>2.741</td>
<td>1.567</td>
</tr>
<tr>
<td>Resistivity (Ω/Squ)</td>
<td>0.086</td>
<td>0.056</td>
<td>0.051</td>
<td>0.069</td>
</tr>
<tr>
<td>Scratch test (% increase)</td>
<td>1.2258</td>
<td>1.3304</td>
<td>1.2076</td>
<td>9.1865</td>
</tr>
<tr>
<td>Mandrel (% increase after 200 cycles)</td>
<td>6.43</td>
<td>3.58</td>
<td>4.27</td>
<td>30.97</td>
</tr>
</tbody>
</table>
Appendix B – Standard Test and Measurement Apparatuses

This section describes standard test apparatuses used in this thesis in order of appearance. The apparatuses shown in this section are shown in the table below.

<table>
<thead>
<tr>
<th>Apparatus</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>Viscosity measurement apparatus</td>
<td>197</td>
</tr>
<tr>
<td>Resistance measurement apparatus</td>
<td>197</td>
</tr>
<tr>
<td>Submersion testing apparatus</td>
<td>197</td>
</tr>
<tr>
<td>Tensile testing apparatus</td>
<td>197</td>
</tr>
<tr>
<td>Mandrel testing apparatus</td>
<td>199</td>
</tr>
<tr>
<td>Biopotential amplification apparatus</td>
<td>200</td>
</tr>
<tr>
<td>Impedance spectroscopy apparatus</td>
<td>202</td>
</tr>
</tbody>
</table>

Viscosity measurement apparatus:

A Brookfield CAP1000+ cone and plate viscometer is used to measure the viscosity of pastes used in this thesis. To measure a paste’s viscosity, the paste is placed on a stationary plate. A cone is then pushed down, sandwiching the paste between the plate and the cone, and turned by the machine at 25 rpm for one minute. The torque is measured by the machine. From this torque the machine can estimate the viscosity of the paste.

Resistance measurement apparatus:

Resistance measurements in this thesis are made using a Tenma 72-7735 multimeter. This measures the resistance between two conductive probes. These probes are placed by hand on the sample being measured. Where the measurement is of a screen printed conductive track, the resistance is measured from the middle of its terminations, whether they are via pads or electrode pads.

Submersion apparatus:

Water absorption is measured using stencil printed samples of each of each material. Tap water is used rather than DI water in order to match domestic washing conditions. The vessel used is a 300 ml beaker and any heating is carried out with a Kern EW3000 hot plate.

Tensile test apparatus:

A Tinius Olsen H25KS is used to test stencil printed material samples to failure. Samples are placed in the machine, held at either end by grips. The samples are stretched at 200 mm/minute and the stress is recorded by the machine. Dumbbell samples are used for all polymer samples while rectangular samples are used for textiles.

Dumbbell samples are stencil printed to produce a dumbbell shaped sample of the material, with a gauge region terminating at either end with a wider gripping part to prevent the material slipping out of the tensile test machine due to dimensional change when stretched.
The gauge region is 10 mm wide, 1.2 mm thick and 25 mm long. The Tinius Olsen equipment and a diagram of the dumbbell sample shape are shown in Figure 9.8-1 and Figure 9.8-2.

![Dumbbell sample diagram](image1.png)

![Tinius Olsen testing machine](image2.png)

**Figure 9.8-1:** Dumbbell sample used in tensile testing. Stencil is 1.2 mm deep.  
**Figure 9.8-2:** Tinius Olsen H25KS tensile testing machine.

The stress on the gauge region is found by dividing the applied force in Newtons by the cross sectional area, which is 0.000012 m². The extension of this region is found by dividing the elongation in millimetres by the gauge length, which is 25 mm. Ultimate tensile strength is the tensile stress when the sample fails, and the Young’s modulus is calculated as shown in Equation 5 and Equation 6.

\[
\text{Young’s modulus at extension } x = \frac{\text{Tensile stress at } x \ (N/m^2)}{\text{extension } x}
\]

**Equation 5:** Young’s modulus at extension x.

\[
\text{Young’s modulus at break} = \frac{\text{Tensile stress at break} \ (N/m^2)}{\text{Extension at break}}
\]

**Equation 6:** Young’s modulus at break.

Several samples are usually used and a mean average is taken for both ultimate tensile strength and elongation at break. Where tensile tests are shown in graphical form, the graph shows only the average while all samples were unbroken. In other words, the plot goes up to the lowest elongation at break. A mandrel test of three samples of Silastic RTV-M with a 12.5 phr carbon loading and the average of these three trials (shown on page 82) are shown in Figure 9.8-3. The vertical line shows the lowest elongation at break of these samples.
Figure 9.8-3: Tensile tests of three dumbbell samples of Silastic RTV-M silicone rubber with a 12.5 phr carbon loading.

Mandrel apparatus:

The mandrel test is used to show the effect of cyclic bending normal to the plane on the resistance of a printed structure. The mandrel apparatus used in this work is shown in Figure 9.8-4.

Figure 9.8-4: Machine mandrel used in comparing the durability of conductive track materials and designs.

It can be adjusted for diameters of 6, 10 and 14mm, but the 6 mm diameter is used throughout this thesis. The mandrel belt has two pockets so that two printed textile samples can be used at the same time. Printed textile samples can be bent either internally or externally, depending on how they are inserted into the mandrel pockets. It has been shown
experimentally that more repeatable results are obtained with longer printed tracks, so printed textile tracks of lengths 50 and 40 mm are used in this work. A DC motor is used so that samples can be bent at a constant speed and force, improving repeatability. The gearing is included because a previous design, that turned the belt directly, had insufficient torque and caused the mandrel’s speed to vary during tests.

Printed conductive track samples are removed from the mandrel pockets periodically, after a set number of cycles around the mandrel radius. Their resistance is measured, and then they are reinserted for further cycles. This process had very high repeatability in terms of its effect on the resistance of a printed conductive track with a given structure and material composition. Three or four conductive tracks were used for each tested structure. As with the tensile tests, graphical representations show the average resistance of a set of samples only while all tracks remain conductive. A mandrel test of four conductive tracks in internal bending and the average of these four trials (shown initially on page 102) are shown in Figure 9.8-5. These conductive tracks were printed on to Escalade, with UoSi-IF-#4 interface and encapsulation layers and a Dupont 5000 conductive layer. The vertical line shows where one of the tracks had ruptured, at 80 cycles.

![Figure 9.8-5: Mandrel tests of four identical printed conductive tracks in internal bending and the average from these trials.](image)

**Biopotential amplification apparatus:**

Biopotentials are amplified using an instrumentation amplifier circuit described by Spinelli et al [74]. The output of the amplifier is connected to the analogue to digital converter input of a National Instruments USB-6008. This allows recording of the ECG signal at 1000Hz into LabView Signal Express 3.0. This data can then be exported to MATLAB or Microsoft Excel for analysis. The standard amplifier circuit and the USB-6008 are shown in Figure 9.8-6 and Figure 9.8-7 respectively.
Four of these amplifiers were fabricated as individual PCBs. R3 and R4 could be altered to change the gain of the amplifier. These four PCBs were wired together into a shielded box that used standard test connectors and shielded cables to connect to printed electrode networks. An X-pole connector was used to power the amplifiers and connect the amplifier outputs to the DAQ device. The shielded box with the amplifiers inside is shown in Figure 9.8-8.

This amplifier is used for all experiments in this thesis except those described in section 7.4, in which active electrodes are examined. In this section an extra amplification stage was added between the electrodes and the main amplifier, aiming to prevent issues caused by the high current of the active electrodes feeding into the amplifiers in the AC coupling stage. It was never shown conclusively that this issue was causing problems with the active electrodes but both active and passive electrodes performed well with the added gain stage. This stage was implemented outside the shielded box on a breadboard and cables from this breadboard to the amplifiers were unshielded. The circuit diagram for the modified amplifier is shown in Figure 9.8-9.
Impedance analysis apparatus:

A Wayne Kerr 6500B impedance analyser is used to measure the impedance of samples fabricated in this thesis over a range of frequencies. This machine connects to the printed samples by crocodile clips attached to the machine’s cables.
Appendix C – Screen Designs

There are eight sets of screen designs in this appendix, an exhaustive list of all screens used in this thesis. The first two sets are comprised of only interface (green) and conductor (red) screens, and the other six are composed of interface, conductor and encapsulation (blue) screens. The sets are listed here in order of appearance in the thesis.

Printed conductive track screens 1 (PC-1):

(a)  
(b)

The first screens used, for interface (a) and conductor (b) layers of conductive tracks with lengths 20-50 mm.

Printed conductive track screens 2 (PC-2):

(a)  
(b)

Screen set PC-1 was revised because conductive track lengths of 40 and 50 mm perform better in a mandrel test. These screens also have tapering rather than corners where the conductive track meets the via pad to improve durability in this area.
Screens for 2 electrodes, tracks and vias (EL-1):

(a) Screens used to fabricate encapsulated conductive tracks with small via pads at one end and larger electrode pads at the other. This was used in the fabrication of a passive electrode chest band to monitor ECG.

(b) Screens for round electrodes with varying diameters (EL-2):

(c) Screens used to fabricate four electrodes with diameters of 30, 20 and 10mm. This was used to examine the effect of contact area with printed passive electrodes.
Active electrode screens (AE-1):

Screens used to fabricate a pair of active electrodes. (a) is the interface layer design, (b) is the conductive layer design, and (c) is the encapsulation layer design. The via pads for power and signal lines to and from the electrodes are on the left, and the electrode and buffer circuit are on the right. A passive, DRL electrode is in the middle.
Frank configuration vest screens (FC-1):

 Screens used to fabricate the Frank configuration vest. (a) is the interface layer design, (b) is the conductive layer design, and (c) is the encapsulation layer design. These screens had dimensions 500 x 500 mm.
Facial EMG/EOG headband screens (HB-1):

(a)

(b)

(c)

Screens used to fabricate the twenty passive electrode network for the facial EMG/EOG headband. (a) is the interface layer design, (b) is the conductive layer design, and (c) is the encapsulation layer design. These designs used the lowest conductive track spacing for any design in this thesis, at 0.5mm with a 0.5mm conductor width. All conductive pads had diameter 10 mm.
Forearm EMG armband screens (AB-1):

(a) is the interface layer design, (b) is the conductive layer design, and (c) is the encapsulation layer design. The conductive via pads had diameter 9 mm and the conductive electrode pads had diameter 11 mm.
Appendix D - Electrode Durability

This section gives a brief discussion of electrode durability. The electrodes developed in this thesis are durable in that they can be used repeatedly for biopotential measurements and none have broken due to usage. Sufficient bonding is achieved with the fabrication process previously described to hold the electrodes in place during experimentation. The electrode systems described are, however, not durable to machine washing, because silicone rubber bonds poorly to other materials and the electrodes would break off during washing. Apart from the poor adhesion, carbon loaded rubber is an ideal, low-cost material for use in such systems so some approaches have been defined to improve the adhesion of electrodes. These methods have not been demonstrated experimentally.

Structural attachment of silicone rubber:

Electrode attachment can also be achieved with structural arrangement of the stencil printed polyurethane. The active electrodes are composed of a carbon loaded rubber electrode printed around a polyurethane cuboid that contains the buffer amplifier circuit as shown in the figure below (a). This polyurethane cuboid is well bonded to the screen printed polyurethane, which in turn is well bonded to the textile. The carbon loaded rubber requires a higher stress to detach from the rest of the printed system because of the stencil printed polyurethane, so this configuration offers greater durability than the passive electrodes, which have no stencil printed polyurethane structure.

![Figure showing structural attachment configurations](image)

The existing active electrode structural configuration (a) and two proposed configurations to increase durability (b and c).

This figure also shows two proposed configurations for improving the durability of the active electrode structure. The figure (b) shows a polyurethane border printed around the carbon loaded rubber, which has a chamfered edge. This has the advantage of preventing water ingress into the electrode, by sealing fully around the edges. However, the creation of this polyurethane border requires an extra fabrication step. It could be implemented with stencil or dispenser printing. Figure (c) shows another proposed configuration. This requires no extra fabrication step, but could not be implemented with the stencil printing technique employed in this thesis because the stencil could not be removed after printing. Some modification to the existing process would be required. Further, there is no prevention of water ingress at the
edges of the electrode. There is no reason why the configurations shown in (b) and (c) could not both be employed, other than the increase to fabrication complexity.

**Reduction of forces detrimental to adhesion:**

During the washing tests of the UoS-IF-#4 and UoS-IF-#39 conductive track structures it was noted that the glob top via encapsulations were more durable than the stencil printed. One possible explanation for this is that the sides of the stencil printed encapsulation were perpendicular to the textile, and so it is more likely that during the random motions in a washing machine that a force can occur that works against the adhesive bond between the stencil and screen printed structures. The current design and the proposed domed shape for electrodes are shown in the figure below.

Printing the carbon loaded rubber in a domed shape would reduce the incidence of forces detrimental to adhesion. This would be difficult to implement with a stencil, and either injection moulding, dispenser printing or 3D printing would be more appropriate.

**Conclusions:**

The main problem with the electrode durability is poor adhesion between the silicone rubber and the screen printed structure. Silicone rubber is known to be difficult to use in a printed structure due to its low surface energy. It has already been shown in this thesis that most pastes cannot be printed on to silicone rubber. However, this material is useful because it vulcanises at low temperatures, can be made conductive, and hundreds of variants are available in a range of viscosities and mechanical properties. Further, some carbon loaded rubbers are biocompatible and approved for medical use however this has not been shown of the formulation described in this work. More work on different types of silicone rubber could yield electrodes that are more conductive and more flexible.

As silicone rubber has poor adhesion, the recommendations here have focussed on mechanically fixing the stencil printed carbon loaded rubber to the screen printed base, and reducing the occurrence of forces detrimental to adhesion. Both of these methods fundamentally aim to reduce forces contrary to the adhesive bond. Another solution would be increasing the strength of the adhesive bond. This would be possible through the use of a conductive adhesion primer to fix the carbon loaded rubber to the screen printed base, or by mechanically bonding it by hooking it into the textile yarns. It is hoped that the designs and
recommendations provided in this appendix will aid in the production of durable electrodes using this technology.