Design and Characterization of Polymeric Strain Gauges for Biomedical Applications

by

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B.ASc, The University of British Columbia, 2012

A THESIS SUBMITTED IN PARTIAL FULFILLMENT OF THE REQUIREMENTS FOR THE DEGREE OF

Master of Applied Science

in

THE FACULTY OF GRADUATE AND POSTDOCTORAL STUDIES
(Electrical and Computer Engineering)

The University of British Columbia
(Vancouver)

March 2015

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Abstract

The market need for organic materials to be used in sensor design has increased with the growing interest in organic printed electronics. Therefore, it is important to find and investigate the piezoelectric and piezoresistive properties of organic materials through the use of alternative rapid fabrication techniques. Poly(3,4-ethylenedioxythiophene) poly(styrenesulfonate), commonly known as PEDOT:PSS, a conductive polymer widely used in organic electronics, can be possibly used as piezoresistive element to measure the strain on flexible substrate electronics. Using PEDOT:PSS and other metallic inks such as silver, the goal of this work is use alternative microfabrication technologies to deposit PEDOT:PSS on flexible substrates and then to use these methods to design strain gauges. The targeted biomedical applications of the designed strain gauges vary from rehabilitation devices to smart biomedical monitoring systems. In this work, PEDOT:PSS strain gauges are initially designed using aerosol jet deposition on a flexible polyamide substrate. The technology has proved to be very powerful in depositing lines with thickness less than 1\(\mu m\). In order to reduce the initial resistance of the strain gauges, it is desirable to increase the thickness of the structure. For this reason, laser micromachining etching is used to fabricate PEDOT:PSS strain gauges. The designed structures have been tested mechanically and electrically in order to measure their gauge factors to longitudinal and transversal mechanical strains. The resultant longitudinal gauge factor varied in the range of -1 and 2, while little change in the resistance was noticed for transversal characterization. Using the same fabrication method, silver paint strain gauges are designed and characterized to have a high longitudinal gauge factor approximated to be higher than 10. The silver paint gauge factor barely responded to transversal actuation. While the variability of the PEDOT:PSS
strain gauges results seemed to be an issue, the reproducibility of silver ink strain gauges proved the viability of the technological fabrication process presented in this work.
Preface

This dissertation is original, unpublished, independent work by the author, M. Al-marghalani in the Adaptive Microsystems Lab at The University of British Columbia (UBC).
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Acknowledgments

I would like to express my sincere gratitude to all the faculty, staff and fellow students who provided gracious support and advice throughout the way. In particular, I would like to thank my supervisor Dr. Edmond Cretu for all the help, advice, emotional and financial support throughout my masters. I also would like to thank Dr. Jean-Sebastien Blouin for hugely contributing mentally and financially to my progress. I am so grateful for Dr. John Madden and Dr. Mu Chiao for allowing me to use their labs equipment. I am very thankful for our lab manager Dr. Alina Kulpa for her assistance and insightful guidance during my work in the lab.

I honestly could not have made it through all of this without the help and support of some particular colleagues and friends in the program: Miguel Torres, Dr. Elie Sarraf, Dr. Mrigank Sharma, Daniel Au, Ahmad Sharkia, Siamak Moori, Carlos Gerrardo, Harrison Brown, Shirely (Lingyi Liu), Saquib Sarwar, Meisam Farajollahi, and Soheyl Kianzad.

I can not express enough my gratitude for my family in Saudi Arabia and my dear friends here in Vancouver and around the world for the emotional support throughout my education. Thank you for believing in me and supporting me no matter what.
To my graciously invaluable family and friends
Chapter 1

Introduction

*I have not failed. I’ve just found 10,000 ways that won’t work.*
— Thomas A. Edison (1931)

The importance of measuring the structural properties of bridges, planes and or even human bodies has paved the way for the design of devices to sense such properties. Strain is just one significant indication of the structural properties and is heavily used in the environment around us. Strain gauges are sensors used to characterize the strain applied on any object from a building to our own muscles. Every muscle in our body incorporates such sensors that transform the strain in a muscle into a signal the brain can understand. The motivation of this thesis to pave the way for the design of strain gauge sensor to be used to associate body strain to electromyogram (EMG) signals produced in muscles. While there are many methods to measure strain, the design and characterization of a novel, yet cheap, material strain gauge is outlined in the thesis.

1.1 Strain Gauges in Biomechanics

Strain is the measure of the relative change in dimensions and shape of an object due to internal or external force. When a certain stretching force is applied to an object, it causes an increase in length of the object due to tensile strain. Also, when the object is compressed, the length of the object will reduce due to a compressive stress along the compression axis. An example of strain in the human body is when
a person uses the index finger to point at an object. The skin of the index finger expands (lengthens) due to the tensile strain applied by index finger muscles. Almost every movement in the human body causes either an eccentric (lengthening) or concentric (shortening) contractions. Those contractions will produce a small electrical signal that is proportional to the neural activity. The signals can then be monitored using an EMG amplifier. While EMG can provide physicians with a lot of information about the muscle activity, the force a muscle produces is dependent on the length of the muscle and the velocity of the contraction, which can not be reflected accurately by EMG alone [47]. Also, EMG amplifiers are expensive, as they have to be certified for human use due to the chance of electrical shock.

Strain gauges are devices that are able to measure strain, or stress for that matter. Typical strain gauges have a spiral-like shape of a metallic wire placed on an insulating stretchable substrate. When the gauge is stretched axially, the dimensions (increase in length in most cases) of the gauge changes and as a result a resistance change is induced. When gauge is stretched transversally, the area (increase in width mostly) of the gauge changes and again, as a result, a resistance change can be measured. While metal foil strain gauges are the most common, some other material strain gauges such as semiconductors or organic polymers are emerging.

Strain measurements can potentially complement EMG measurements to pave the way for many biomedical applications including haptic interfaces and smart prostheses.

### 1.1.1 Piezoresistivity and Gauge Factor

As mentioned before, the change in the mechanical dimensions causes a change in the electrical resistance in strain gauges. This effect is known as piezoresistivity. The first part of the word *piezo* comes from Greek, which means press or compress. If we consider a long wire with cross section $A$, length $l$ and resistivity $\rho$ as shown in Fig. 1.1, the electrical resistance $R$ can be calculated as

$$R = \rho \frac{l}{A}$$  \hspace{1cm} (1.1)
\[ R = \rho \frac{l}{\pi r^2} \]  

(1.2)

**Figure 1.1:** A cross section of a long wire showing the resistivity, cross section and length, adapted from [6]

A change in resistance divided by the initial resistance results in the following

\[ \frac{\Delta R}{R} = \frac{\Delta l}{l} - \frac{2\Delta r}{r} + \frac{\Delta \rho}{\rho} \]  

(1.3)

Now we can use the Poisson’s ratio \( v \), defined as the negative ratio of the strain in the transversal \( \varepsilon_t \) direction to the strain in the longitudinal \( \varepsilon_l \) direction, as

\[ v = -\frac{\varepsilon_t}{\varepsilon_l} \]  

(1.4)

or

\[ \frac{\Delta r}{r} = -v \frac{\Delta l}{l} \]  

(1.5)

As a result, we can write Eq. 1.3 as shown

\[ \frac{\Delta R}{R} = (1 + 2v) \frac{\Delta l}{l} + \frac{\Delta \rho}{\rho} = (1 + 2v + \pi E) \varepsilon \]  

(1.6)

where \( \pi \) is the piezoresistive coefficient. Eq. 1.3 can be also written as

\[ \frac{\Delta R}{R} = (\frac{\Delta R}{R})_l + (\frac{\Delta R}{R})_t = GF_l \varepsilon_l + GF_t \varepsilon_t \]  

(1.7)

where \( GF_l, GF_t, \varepsilon_l, \varepsilon_t \) are the longitudinal gauge factor, transversal gauge factor, longitudinal strain, and transversal strain, respectively.

In metal foil strain gauges, the term \((1 + 2v)\) in Equation 1.6 represents the geometrical effect and is dominant, as the change in intrinsic piezoresistivity term
Table 1.1: Gauge factors of some metals [2].

<table>
<thead>
<tr>
<th>Metal</th>
<th>Gauge Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>Copper</td>
<td>2.2</td>
</tr>
<tr>
<td>Constantan</td>
<td>1.9</td>
</tr>
<tr>
<td>Nickel</td>
<td>2.7</td>
</tr>
<tr>
<td>Platinum</td>
<td>2.4</td>
</tr>
<tr>
<td>40% gold/palladium</td>
<td>1.9</td>
</tr>
</tbody>
</table>

$\frac{\Delta \rho}{\rho}$ is small [12]. When a metal is pressed, the change of the mobility in electrons due to the volume change will cause a small change added to the geometrical effect of the gauge. Hence, typical values of the gauge factor of metals are positive and in the range of $1 - 2.7$. Examples of gauge factor values of some metals represented in Table 4.1.

For an isotropic material, the piezo resistive effect is given by

$$\frac{\Delta \rho_x}{\rho_x} = \pi_x \sigma_x$$

where $\sigma_x$ is the stress and is related to the strain by Young’s modulus $E$ according to Hook’s law

$$\sigma = E \varepsilon$$

In order to describe the effects of stress of all dimensions, including normal and shear stress of an anisotropic material, the piezoresistivity coefficients must be a tensor such that

$$\frac{\Delta \rho_i}{\rho_0} = \sum_j \pi_{ij} \sigma_j$$

In Equation 1.10 the index $j$ of the stress term $\sigma$ encompasses both the three normal and three shear components of stress. Similarly, the relationship of voltage and current as expressed in Ohm’s law depends on the direction of the electric field and has 6 components reflected in the index $i$. In a matrix form, Equation 1.10 can be
The number of independent piezoresistive components represented in the matrix can be reduced for anisotropic materials due to symmetry. Silicon crystals have only 12 non-zero elements in the matrix and only three independent components, which are: \(\pi_{11}, \pi_{12}\) and \(\pi_{44}\). Only the piezoresistive coefficients in the longitudinal, which is parallel to the direction of current flow, and transversal, which is perpendicular to the current flow, are considered in most applications [25]. In semiconductor based strain gauges, the term \(\Delta \rho / \rho_0\) is much more dominant than the geometrical effect. Increasing the gauge factor of strain gauges is very desirable as the gauge factor represents the sensitivity of the sensor. Due to the high intrinsic piezoresistivity of semiconductor strain gauges, the gauge factor of such devices is approximately two order of magnitude higher than metallic gauges [42] [19]. While semiconductor strain gauges have a relatively high gauge factor in comparison to metallic foil gauges, they have much greater sensitivity to temperature-making them less reliable in applications where temperature variation is expected. Add to that, semiconductor strain gauges have a nonlinear relationship of the stress to resistance, but this drawback can be easily dealt with in today’s advanced computer systems [12].

### 1.1.2 Electromyography (EMG) and Strain Measurements

Electromyography (EMG) is a valuable technique to analyse and understand the electrical activity in a muscle or a group of muscles. The EMG signal is generated in the muscle fibres during a contraction of the muscle. The actual source of the electrical signal comes from the depolarization process that happens in the muscle during a contraction and is separated from the recording electrode by layers of
tissues [29]. As mentioned earlier, while EMG can provide practitioners with a lot of information about the muscle activity, EMG alone does not reflect the muscle force precisely. Tension measurements, in addition, can possibly complement the information given by EMG about the muscle force [47]. Some studies have demonstrated the potential of using strain measurements and correlating the results with EMG data [47] [48]. In 1952, a study of the relationship between EMG and tension of the bicep brachii in humans indicated a parallelism of the tension to the integrated EMG signal [18]. While the first study used a dynamometer to measure the tension produced by the isometric contraction, it did not incorporate a strain gauge on site of measurement. Interest in measuring selectively the force of muscles in humans or animals led to implantation of strain transducers in animals at first. A study that used implantation of an EMG amplifier and a mercury-based strain gauge in rats demonstrated clearly a direct relationship of strain measurements and integrated EMG signal during lordosis reflex in rats [36]. However, EMG alone can not provide precise information, as it is the sum of motor units potentials, while the generated force also depends on the muscle size, and the firing rate [48]. Precise measurements of force generated in muscles and tendons led to surgically planting a strain gauge in a human achilles tendon for accurate force measurements. The implantation of such sensors was made possible by following the same procedure done in cats earlier. The purpose of the study was to investigate clearly the role of the achilles tendon in jumps and to study directly the elastic force generated and that was possible through the data provided by the implanted metallic foil strain gauge [14] [13].

While implantation of devices in humans can provide direct and accurate measurements, this method is cumbersome due to its invasiveness. A group in Slovenia has developed a strain gauge based sensor to selectively and non-invasively measure the muscle force [47] [48]. The development of high sensitivity and inexpensive strain gauges promises a bright future in non-invasive biomechanics measurements. The designed sensor setup and the results of the strain sensing of the bicep brachii from [47] is shown in Fig. 1.2 and Fig. 1.3, respectively.
Figure 1.2: A simplified representation of the muscle contraction sensor (MC) measuring principle for the determination of the mechanical and physiological properties of skeletal muscles (1): Sensor tip; (2): Skin and intermediate layer; (3): Measured muscle, adapted from [47]

Figure 1.3: Simultaneously recorded force from the Muscle Contraction sensor $F_{MC}$, force from a dynamometer $F_D$ and EMG during isometric contraction of biceps brachii muscle of subject 3 (n3), adapted from [47]
1.2 Conductive Polymers

As the demand for flexible organic electronics has been increasing in the past few years, there is a need for investigating organic materials that have the potential to be used in organic sensor designs. Most of these materials embrace conductive properties that qualified them to substitute conventional metals in some applications. Organic light emitting diodes, organic solar cells and organic transistors are some of the key developments using conductive polymers [35] [43] [45].

The field of organic electronics has firstly seen the light in middle of the twentieth century. Development in the field has led 3 researchers, Alan J. Heeger, Alan G. MacDiarmid and Hideki Shirakawa, to win the Nobel prize in chemistry "for the discovery and development of conductive polymers" [41].

Conjugated polymers (conductive polymers) have long chains of carbon with single and double bonds alternating. The extended π-orbital system in the polymer allows electrons (or holes) to move from one end of the chain to the other [9] [34].

Table 1.2 shows some examples of conjugated polymers, their bandgap energy and highest reported conductivities. It should be noted that the bandgap energy of silicon is 1.1eV. It can realized that the availability of p-orbitals is an important factor for the polymer to be intrinsically conductive, as this will provide an orbital system for the charge carries to move along the polymer [9].
Table 1.2: Some conjugated conducting polymers, their bandgap energy and conductivities, adapted from [9].

<table>
<thead>
<tr>
<th>Polymer (date conductivity discovered)</th>
<th>Structure</th>
<th>MM gap (eV)</th>
<th>Conductivity* (S/cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>I. Polyaniline and analogues (1977)</td>
<td></td>
<td>1.5</td>
<td>10^5 – 1.7x10^5</td>
</tr>
<tr>
<td>Poly(pyrrole) (1979)</td>
<td></td>
<td>3.1</td>
<td>10^2 – 7.5x10^2</td>
</tr>
<tr>
<td>Polythiophene (1981)</td>
<td></td>
<td>2.0</td>
<td>10 – 10^3</td>
</tr>
<tr>
<td>II. Polyphenylene and analogues (1979)</td>
<td></td>
<td>3.0</td>
<td>10^3 – 10^4</td>
</tr>
<tr>
<td>Poly(p-phenylene vinylene) (1979)</td>
<td></td>
<td>2.5</td>
<td>3 – 5x10^1</td>
</tr>
<tr>
<td>Polyline (1980)</td>
<td></td>
<td>3.2</td>
<td>30 – 200</td>
</tr>
</tbody>
</table>

Organic conductors posses a few advantages over conventional materials. The advancement of such materials has been built upon the affordability and the flexibility of the conductive polymers [37]. Also, since the chemical structure of such materials can be easily modified, a range of properties of the material can be precisely tuned. For example, the doping of the polymer determines the conductivity of the material. While organic polymers are not normally conductive by nature, the process of transforming a polymer incorporates either partial oxidation with electron acceptors, such as AsF₅, or partial reduction with electron donors, such as Na. While the doping process to transform a polymer from an insulator to a conductor is not simple, it adds the tunability to the conductivity of the material [9] [15]. Altering the chemical structure of a conductive polymer can also add bio-compatibility to its properties, which make conjugated materials favorable in biomedical applications [37].

Another important property of conjugated polymers is the ability to dilute the polymer solution for better transparency, a feature that is highly desirable in optoelectronics applications. As a result, there has been a lot of interest in the use of conjugated polymers in light emitting diodes and touch screen applications [37] [35] [8].
While there are so many advantages of conductive polymers, there are still some challenges in the development of such materials. The lifetime expectancy of organic materials after being exposed to environmental aspects poses the biggest challenge in commercializing organic electronics devices. Packaging and encapsulation of organic devices can provide a promise to the evolution of organic devices. Another drawback of conjugated materials lies in the performance of such devices in comparison to the mature metal or semiconductor devices. While the performance is lagging behind the mature microelectronics technology, organic electronics can still complement the market with cheap and disposable devices [24] [37].

1.2.1 PEDOT:PSS

Poly(3,4-ethylenedioxythiophene) poly(styrenesulfonate), also commonly known as PEDOT:PSS, is an intrinsically complex conductive polymer widely used in organic electronics and organic photovoltaic (OPVs). PEDOT is one of the derivatives of thiophenes and is a conjugated polymer extracted from ethylenedioxythiophene (EDOT) monomers in the reaction shown in Fig. 1.4 [38]. While PEDOT is the conductive part of the solution, it is normally doped with PSS in order to increase its solubility. [10]. Other dopants used with PEDOT including Tosylate and phosphomolybdate [37]. The conductivity of PEDOT:PSS lies in a net positive charge on the PEDOT chain that attracts the negative charge left on the

![Figure 1.4: Chemical structure of PEDOT:PSS post polymerization of EDOT in polystyrene sulfonic acid, adapted from [38]](image)

While there are so many advantages of conductive polymers, there are still some challenges in the development of such materials. The lifetime expectancy of organic materials after being exposed to environmental aspects poses the biggest challenge in commercializing organic electronics devices. Packaging and encapsulation of organic devices can provide a promise to the evolution of organic devices. Another drawback of conjugated materials lies in the performance of such devices in comparison to the mature metal or semiconductor devices. While the performance is lagging behind the mature microelectronics technology, organic electronics can still complement the market with cheap and disposable devices [24] [37].

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acids. As a result, PEDOT and PSS are closely connected the attraction between the left positive charges on the PEDOT chains attract the negative charges on the PSS chains. Once the PEDOT:PSS compound is bonded, an unpaired pi electron is highly mobile on the chain leading to high conductivity. PEDOT:PSS is available commercially under the trade name Clevios™ from Heraeus or Orgacon™ from Agfa. The polymer was initially used as a preventing layer for static charge in photographic films. Agfa coats about 200 million films a year with a thin layer of PEDOT:PSS to prevent electrostatic discharge during the production of films [37].

PEDOT:PSS has many advantages in comparison to other types of synthetic polymers. The low oxidation potential and the moderate bandgap of PEDOT:PSS make the polymer more attractive for organic electronics applications. The bandgap of PEDOT:PSS can vary, depending on the preparation of the solution, in the range of 1.6eV to 2.5eV, with a work function in the range of 5.2eV [38] [11]. While environmental stability of doped PEDOT is a key feature when compared to other conducting polymers, it can also be transparent with relatively high conductivity-making it an economically beneficial choice for optoelectronics applications.

Due to the flexibility of PEDOT:PSS, many attempts have been made to design electromechanical sensors using this material. Strain sensors based on conductive polymers can resist a high strain range, higher than metals and semiconductors, while still being relatively conductive [39].

Fabrication of PEDOT:PSS sensors are so far based on traditional clean-room processes. Photolithography was used by Mateui et al. to produce a PEDOT strain gauge sensor on silicon substrate [28]. The use of traditional fabrication methods can hinder the development of such technology, as it requires long processing time and higher costs. Other alternative fabrication methods such as slot die coating, spin coating, inkjet or screen printing were reported [1] [38].

1.2.2 PEDOT:PSS Based Stress Sensors

Piezoresistivity and various patterning techniques of PEDOT:PSS have been reported in the literature. PEDOT:PSS can be patterned using conventional clean-room photolithography and polymerization processes, but this could be problematic as the fabrication process is expensive for small quantities of sensors. To the
author’s knowledge, the first attempt to characterize the piezoresistivity of PEDOT:PSS used spin coating techniques to deposit a thin layer of PEDOT:PSS on Kapton polyamide film. The layer deposited thickness was approximately 50 nm at 4000 rpm. After the deposition of the PEDOT:PSS layer, gold or silver contacts were deposited using shadow masking. The final device structure is shown in Fig. 1.5.

![Figure 1.5: The device structure of the PEDOT:PSS strain gauge, adapted from [37]](image)

After characterizing the longitudinal piezoresistive coefficient, they concluded that the measurable resistance is partially cancelled out by the geometrical effect, resulting in a gauge factor in the range of $-1.1$ to $0.3$ [37]. One of the main issues of this work was the initial resistance of the device was in the range of Mohm, making it hard to detect accurately the change in resistance.

A group from Technical University of Denmark fabricated PEDOT strain gauges using conventional UV-lithography and reactive ion etching (RIE) processes and reported a gauge factor of $3.41 \pm 0.42$. The group showed the gauge factor of PEDOT is comparable to metal and thus the polymer can be used in all polymer MEMS-based devices [28]. In 2008, another group presented a lift-off process for depositing thin film PEDOT:PSS and showed a proof-of-concept test results of piezoresistivity of PEDOT:PSS [20]. They showed a new fabrication technique of PEDOT:PSS incorporating lift-off as illustrated in Fig 1.6.
The method consists of depositing polyamide on a silicon wafer followed by patterning photoresist- exposing the areas of interest for the PEDOT:PSS deposition using spin coating. After PEDOT:PSS is deposited, the wafer if flipped in (d) and another layer of photoresist is deposited for dry etching under the areas of PEDOT:PSS. They then characterized the pressure gauges designed and reported a gauge factor of $0.48 \pm 0.07$ at $36.6 \text{ pm}^3$ relative humidity [21].

PEDOT:PSS was fabricated elsewhere using a peeling technique, which incorporates clean room processes such as etching and spin coating, and then characterized using micro-bending. The results of the bending showed a high gauge factor of $17.8 \pm 4$, which is well above aforementioned results of PEDOT:PSS and other metallic strain gauges [23].

The range of operation and the reliability are two important factors in strain gauges. As shown in [39], a flexible foam structure was used as a substrate for the PEDOT:PSS to study the reliability of the material. A thin layer of adhesive polyurethane (PU) was deposited at first to make the porous foam hydrophobic followed by spin coating of PEDOT:PSS. Laser micromachining was then used to pattern the structure. It was shown that PEDOT:PSS can withstand a displacement of $17.7\%$ for 60 cycles, which is much higher than the range of operation of metallic and semiconductor strain gauges.

PEDOT:PSS has been of interest for large area tactile sensors to be used in robotics applications. The main reason for choosing this polymer is the cost of production drops significantly when compared to metals such as silver. Also, the
flexibility and ease of processing of PEDOT:PSS contribute significantly to this polymer to be the first choice for new generations of sensors. In [5], PEDOT:PSS sheet was deposited using spin coating on an array of electrodes on a printed circuit board (PCB) as shown in Fig. 1.7. When a pressure is applied on the array, the contact between the electrodes and the sheet increases-causing a detectable change in resistance.

Figure 1.7: PEDOT:PSS thin film deposited on an array of electrodes on a PCB, adapted from [5]

The flexibility of PEDOT:PSS has led to integrating the polymer as a sensing piezoresistive sensor in clothing fabrics. Inkjet printing of PEDOT:PSS into the fabrics were shown to cover individual fabrics to the yarn throughout the entire thickness of the cloth. The experiment concluded that PEDOT:PSS has a negative value gauge factor in the range of −5 to −20, which compares to conventional strain gauges. Preliminary data of PEDOT:PSS incorporated in fabrics sensors show that such sensors can be used to monitor knee and wrist motions, which promises to be the future for rehabilitation applications [4] [7].

The use of PEDOT:PSS mixed with other materials have also been reported in the literature. The aim of doing so is to improve either the mechanical properties of the blend or to add conductivity to it. In a paper characterizing the mechanical properties of PEDOT:PSS mixed with Polyvinyl alcohol (PVA), it was shown that adding PEDOT:PSS to the mixture improves the Young’s modulus of PVA from 0.0412GPa to 1.65GPa (50 wt% of PEDOT:PSS). While mixing PE-
DOT:PSS with PVA improved the tensile strength of the nanofibres, the conductivity was hindered. The study suggested addition of 30%-40% of PEDOT:PSS for a reasonable trade off of the mechanical to electrical properties [16]. Another study incorporated this mixture in a design of a strain gauge sensor by electrospinning of PEDOT:PSS/PVA on a flexible substrate. The device demonstrated in the paper exhibited high sensitivity to very small deformation, fast response and a gauge factor up to 396 [26].

The trend in sensor technology tends to move towards all polymer integrated sensors. The addition of PEDOT:PSS as either part of a mixture or as an electrode material has been reported in other studies. In [40], PEDOT:PSS electrodes sandwiched porous Polyniline doped with camphorsulfonic acid (PANI:HCSA). When external pressure is applied on the device, the resistance reduces due to the increase in the contact between the PEDOT:PSS electrodes through the porous mixture.

While many attempts in characterizing PEDOT:PSS were taken in the literature, the development of PEDOT:PSS strain gauges using cheap and rapid alternative fabrication techniques has not yet been explored and evaluated appropriately.

1.3 Motivation

The development of a conjugated polymer strain gauge has been the interest in research for the past decade. Applications varying from skin-mountable strain measurements to structural monitoring require flexible, yet sensitive sensors. Interest in monitoring joint movements in human has led to designing highly stretchable strain gauges [27]. Also incorporation of such devices into fabrics for applications such as rehabilitation paved the way for the development of smart textiles [4]. While the flexibility of the material can be a determining factor is some biomedical applications, less flexibility with higher sensitivity have other applications in the field. As mentioned earlier, PEDOT:PSS has many advantages over other synthetic polymers due its relatively high conductivity, flexibility and transparency.

Metal foil strain gauges have been used mainly for structural monitoring. Two of the important requirements for biomedical strain sensing applications are: flexibility and biocompatibility. Metallic foil gauges have the ability to only measure micro strain- not a suitable choice for joint measurements as the range of strain is
large [30]. However, customized printable metallic based strain gauges can possibly be used in some biomedical applications not requiring a high range of operation. Semiconductor strain gauges have a large gauge factor, but the nonlinearity and the flexibility of the sensors present a problem when considered in biomedical applications.

The PEDOT:PSS conduction mechanism in the sense that the alignment in the chains might contributes to its piezoresistivity. That is, when the polymer is stretched, the alignment of chains can get distorted, leading to a change in the resistivity of the material. While many attempts where taken to develop PEDOT:PSS strain gauges, commercialization of such devices require more research. Many of the developed PEDOT:PSS strain gauges reported seem to have different gauge factors. The reported longitudinal gauge factors for PEDOT:PSS strain gauges range from $-20$ to $17.8$ depending on the fabrication and the characterization methods. Also, no data available in the literature regarding the transversal gauge factor of PEDOT:PSS. Extracting the longitudinal and transversal gauge factors of PEDOT:PSS is a very important step towards commercializing PEDOT:PSS based strain gauges.

The main goal of the thesis is to explore alternative microfabrication methods for the patterning of conductive layers (polymers or metallic) on flexible substrates. Strain gauge designs have been used as target, to explore the spread claims regarding the piezoresistivity of PEDOT:PSS, and to compare polymer and metallic ink based strain gauges.

1.4 Previous Work

This thesis is a continuation of work started initially by developing PEDOT:PSS strain gauges using inkjet printing. In [1], the initial goal was to develop cheap strain gauges based on silver nanoparticles inks. Throughout the work, the relatively inexpensive PEDOT:PSS ink replaced the silver nanoparticles. The thesis work showed the possibility to fabricate strain sensors based on cheap materials using nonconventional cleanroom processes. While the thesis work has clearly showed the potential of using inkjet printing technology for patterning strain gauges, it identified problems with using this printing method. Limited viscosity of the inks, compatibility of the ink with the printing methods, and the
roughness of the printing surfaces were some of the challenges and limitations found in the previous work. While it was reported that PEDOT:PSS does not hold piezoresistive properties after inkjet printing, an experiment setup was designed to extensively study the piezoresistivity of PEDOT:PSS. The thesis work reported a gauge factor of a PEDOT:PSS patterned sensor on Kapton to be 3.63 [1].

The direction of the work switched from using PEDOT:PSS as the base material to synthesizing evaporation cast thin film carbon nanotubes (CNT) as a base material for strain gauges. In [3], the motivation of the thesis was to design CNT based strain gauges for biological and structural health monitoring. CNT have been reported to have high gauge factors, in the range of 600 to 1000, due to the intrinsic piezoresistivity [31]. In this thesis work, CNT were incorporated into various evaporation cast films in order to align the tubes during evaporation. Inkjet printing with air flow evaporation casting were used in hope to achieve alignment of the CNTs. PEDOT:PSS was used as a conductive medium to the CNT film to the vacant space of the substrate. Initial deposition of PEDOT:PSS on the CNT film resulted in displacement of the CNT ink as shown in Fig. 1.8. As a result, it was essential to use an inkjet printer to deposit small and controlled amount of PEDOT:PSS that would sit on top of the CNT film.

![Figure 1.8: PEDOT ink deposition images; evaporation cast PEDOT ink resulting in CNT film displacement on the left and inkjet printed PEDOT resulting in fine lines and absence of CNT film displacement on the right, adapted from [3]](image)

Tensile measurements of the CNT film was performed and resulted in a gauge
factor ranging from 0.1 to 4. The considerably low value of the gauge factor was believed to be due to the lack of alignment of the CNTs embedded into the thin film.

1.5 Chapters Overview

The thesis is divided into four chapters outlining the fabrication, characterization, results along with a discussion, and conclusions with future outlook. The next chapter introduces the microfabrication techniques used to make the PEDOT:PSS strain gauges. It also includes a brief comparison of 3 differently fabricated PEDOT:PSS sensors. The characterization chapter focuses on the methods undertaken to design the characterization setup including white light interferometry and tensile testing. After establishing the ground of characterization, the results chapter shows the obtained data with more analysis of the results. A brief comparison of the results of the different fabrication techniques used is outlined then. Finally, conclusions are presented of the work done in this thesis with some future outlook and potential ideas to improve the design of such sensors.
Chapter 2

Fabrication

In order to extract the piezoresistive coefficients of PEDOT:PSS, different fabrication technologies were investigated to prepare PEDOT:PSS strain gauges. Flexible polymer based substrates were used to prepare the structure for the experiment. The design of the sensor was done by either aerosol jet printing or laser micromachining. Silver paint and copper tape were used to form metal contacts to the resistive structures. The final design of the PEDOT:PSS-based devices is shown in Fig. 2.1. Laser micromachining etching is also used to design silver ink-based strain gauges.

Figure 2.1: The final structure of the strain gauge fabricated using: (a) Aerosol jet Printing (b) Peel off and Laser Micromachining (c) Etching by Laser Micromachining
2.1 Substrate

Two different substrate materials were considered in the design of the PEDOT:PSS strain gauges. The first choice was polyamide, or Kapton®, from DuPont. Polyamide is well studied in the literature and it has very robust mechanical properties—making it an excellent choice for a straining application [32]. Kapton® is used in applications requiring a large range of temperature stability, ranging from $-269^\circ C$ to $400^\circ C$. The stress-strain relationship and the elastic modulus of the Kapton tape were taken from the datasheet from DuPont website.

PEDOT:PSS strain gauges were also designed on the high strength elastomer bonding tape, VHB™ from 3M™. The reason of considering the VHB tape is the high flexibility it provides—making it extremely easy to stretch the PEDOT:PSS sensor. The elastomer has been reported to have a range of flexibility exceeding 100% of its original length [17]. The elastomer also has layer of adhesion, but after a few experiments it was noticed that PEDOT:PSS films do not adhere to the tape. The other issue with VHB is that it is viscoelastic—giving it hysteresis properties that are not desired in strain gauges. Nonetheless, VHB was still used as a substrate material for some samples as it will be mounted on a stiffer material, which forces the VHB to get back to its original length after stretching. Silver ink-based strain gauges were only deposited in Kapton since the stretchability of VHB is too high for a metallic material.

2.2 Fabrication Methods of the PEDOT:PSS Strain Gauges

Three alternative to cleanroom fabrication methods were adapted to design the PEDOT:PSS strain gauges, namely: Aerosol jet printing, laser micromachining and peeling-off, and laser micromachining etching. Each of the fabrication techniques is explained in detail in the following sections.

2.2.1 Deposition Using Aerosol jet Printing

In the fabrication of the PEDOT:PSS strain gauges, the Optomech Aerosol jet printer (Optomech, USA) was initially used to pattern PEDOT:PSS, Agfa Orgaco™
grade IJ-1005, purchased from Sigma Aldrich, lines. Silver paint (SPI, USA) is then used to make electrical pads to connect the PEDOT:PSS lines to copper wires for further measurements. The principle of operation of the aerosol jet printer lies in atomizing the PEDOT:PSS ink using an ultrasonic atomizer, creating a dense aerosol composed of droplets. The aerosol is then transported to the deposition head using a carrier gas. The sheath gas is then used to focus the aerosol on the substrate while the deposition head is patterning the structure. This process will result in the desired structure printed on the substrate with a relatively high width resolution. Kapton polyamide tape (DuPont, USA) is used as a substrate due its flexibility and usability for printed electronics as mentioned earlier. The substrate is initially mounted on a glass slide to allow reasonable flatness during the printing process. The polyamide is cleaned using isopropanol and left to dry for 15 minutes. In order to reduce the viscosity of the PEDOT:PSS before atomizing, the material was mixed with deionized water with a 1:1 ratio, making it easier to atomize using the ultrasonic atomizer. Once the PEDOT:PSS lines are patterned on the polyamide, the structure is left to dry at room temperature for 30 minutes. Silver paint is then applied and the structure was baked at 60°C for 15 minutes. A flow chart of the process is shown in Fig. 2.2.
Although the line width resolution of the aerosol jet printer is high, the width of the PEDOT:PSS lines patterned on the polyamide were increased to 400um in order to decrease the initial resistance of the structure. The geometry of the structure is shown in Fig. 2.3.

2.2.2 Laser Micromachining and Peel-off

The design process of the PEDOT:PSS strain gauge incorporates laser micromachining of a plastic mounted on a VHB elastomer layer. First, a small piece of VHB is put on the adhesive side of Kapton. Then, laser micromachining cutting (Oxford Lasers, UK) is tuned to 355\textit{nm} wavelength, 0.3\textit{mm/s} translation platform speed to output 400\textit{Hz} pulses of 0.12\textit{mJ} energy in order to cut the red plastic (mask) covering the VHB layer into the shape of the sensor as shown in Fig. 2.4.
Figure 2.3: Dimensions of the PEDOT:PSS strain gauge designed using aerosol jet printing technology

Figure 2.4: Exposed VHB layer for PEDOT:PSS deposition after laser micromachining.

Once the mask is cut, the sensor area is removed, exposing the VHB substrate, and then 1mL of PEDOT:PSS is deposited on the mask, as shown in Fig. 2.5, using a regular lab syringe.
Figure 2.5: Deposition of PEDOT:PSS ink on the exposed area of VHB using a regular lab syringe.

Afterwards, the structure is placed in a furnace for 3 hours at 60°C to evaporate the solvent in the PEDOT:PSS and left to anneal slowly to room temperature. Due to the poor adhesion of the PEDOT:PSS to the VHB substrate, peeling off the plastic will cause some parts of the sensor to be come off. Therefore, another run of laser micromachining at a much less laser intensity (355nm wavelength, 1mm/s translation speed, 400Hz pulses and 0.024mJ), is required to separate parts of the PEDOT:PSS from plastic mask. Now the mask is removed, silver paint and copper tape are used to probe the structure at 4 points, each of which is placed on a corner of the structure. Finally, another layer of Kapton Tape is put on top of the structure, making the PEDOT:PSS layer sandwiched in between VHB and the Kapton. Fig. 2.6 (a) shows a side view of the layers in the sensor and (b) the dimensions as viewed from the top.

Figure 2.6: (a) A side view of layers of the sensor incorporating VHB as a substrate. (b) A top view of the PEDOT:PSS dimensions.
The process of laser machining and peeling-off flowchart is shown in Fig. 2.7.

![Flowchart of the process of designing PEDOT:PSS strain gauges using laser machining and peeling-off.](image)

**Figure 2.7:** Flowchart of the process of designing PEDOT:PSS strain gauges using laser machining and peeling-off.

### 2.2.3 Laser Micromachining Etching

Another approach to design ink-based strain gauges is to incorporate laser micromachining etching. This method was considered due to some disadvantages of the VHB elastomer, discussed later in this thesis. For PEDOT:PSS-based gauges, the design process starts with depositing 3 mL of PEDOT:PSS ink on the adhesive side of Kapton tape in an area of 8 mm height × 11 mm width as shown in Fig. 2.8.
Figure 2.8: Deposition of PEDOT:PSS on Kapton using a regular lab syringe.

The structure is then placed in a furnace oven set at 60°C for 3 hours and annealed slowly to room temperature. After that, the laser was set to 355nm wavelength, 0.3mm/s translation platform speed to output 400Hz pulses of 0.12mJ energy to etch out all the unwanted areas of PEDOT:PSS. Again, silver paint and copper tape are used afterwards to electrically connect 4 electrodes to the sensor. Finally, another layer of Kapton is placed on top of the structure to ensure that the force is transformed between the Kapton layers to the PEDOT:PSS structure. A side view of the structure is shown in Fig. 2.9.

Figure 2.9: A side view of layers of the sensor incorporating Kapton as a substrate

The process of laser itching flowchart is shown in Fig. 2.10.
Figure 2.10: Flowchart of the process of designing PEDOT:PSS strain gauges using laser machining etching.

For the silver ink gauges, the process is almost the same, excluding the annealing process. The silver ink used dried at room temperature without the need of furnace. This is advantageous since the throughput of silver ink gauges is higher.
Chapter 3

Characterization

After the fabrication process of the PEDOT:PSS strain gauges sensors, it is important to understand the characterization methods required to identify the physical properties of the design. First, white light interferometry techniques are used to extract the topography of the structure. Then, tensile testing is used to study the effect of the mechanical strain on the electrical resistance of the sensors. In order to perform the mechanical testing of the strain gauges, a design of a beam for the sensors to be mounted on was required. This chapter covers the principles of white light interferometry, tensile testing, and beam design.

3.1 White Light Interferometry

In order to characterize the topography of the PEDOT:PSS structure, Polytec MSA-500 Micro System Analyzer is used. The topography measurement the system incorporates is based on white light interferometry with high speed electronics and software for image processing. The principle of operation is basically splitting a light source into 2 parts, as shown in Fig. 3.1, the first of which goes to a reference mirror, and the other is incident to the test object [44].
Figure 3.1: The principle of operation of white light interferometry, adapted from [33]

When the distance changes between the sample and the interferometer, a change in phase of the reflected wave off the object changes at every depth of the sample in comparison with the reference beam. During the interference scan, a video camera is recording the interference patterns. Then, the software processes the video to acquire the topography of the structure.

3.2 Tensile Testing

In order to accurately characterize the designed strain gauges, simultaneous electromechanical tensile testing is required. The Bose ElectroForce® tensilometer (Bose, USA) can either control accurately the displacement or the force applied to the test object, while simultaneously reading electrical signals. The setup of the experiment using the tensilometer available in Dr. Madden’s lab is shown in Fig. 3.2.
Figure 3.2: Tensile testing setup. 4-point measurement technique is used by supplying the current and measuring the voltage produced by the strain gauge mounted on an aluminium beam.

Initially, the strain gauges are clamped directly in the tensilometer. The test is done by controlling the load while reading the displacement. However, due to hysteresis issues, it was thought that mounting the strain gauges on a material with a high modulus of elasticity and low hysteresis profile can provide better measurements. Therefore, a design of an aluminium beam was required and is discussed in detail in the following section.

In order to avoid contact resistance in the measurements, a 4-point measurement technique is incorporated. Electrical current was applied to the structure using Keithley model 6221 current source (Keithley, USA) while measuring the voltage drop in the strain gauge resistance. The method avoids the contact resistances, as the input impedance to the voltmeter (tensilometer) is very high, making the current flowing through the least resistive path (the strain gauge in this case) as shown in Fig 3.3.
3.3 Beam Design

As mentioned earlier, due to the hysteresis of the substrate materials and some slippage in the clamping of the sensor, mounting the strain gauges on a stiffer beam with a low hysteresis is essential for precise measurements. While the dimensions of the beam were determined by the physical dimensions of the tensilometer, it is of great importance to understand the areas of stress exhibited on the beam. For this reason, analytical calculations of the beam followed by finite element modelling were carried out.

3.3.1 Analytical Model

Consider a beam as shown in Figure 3.4. Since the value of interest of the stress is located on the top surface as the sensor will be mounted there, a computation of the strain on the surface is required. Assumptions made here are:
(i) perfect adhesion between the substrate of the strain gauge and the cantilever beam is made in this computation. That is, the strain of the cantilever beam is perfectly transmitted to the strain gauge. If imperfect adhesion occurred, the resultant strain can be multiplied with $1 - \alpha$, where $\alpha$ is a loss factor.

(ii) the presence of the strain gauge on the surface of the beam does not influence the elasticity of the cantilever beam, as the dimensions of the strain sensor are much smaller than the beam.

From (i), when the cantilever bends, the strain sensor (resistor and substrate) will bend with the same radius of curvature $r$ in mm, and for a small curvature:

$$\kappa = \frac{1}{r} = \frac{dx}{d\phi}$$  \hspace{1cm} (3.1)

The natural axis is at $y = 0$ and $\epsilon_x = 0$. The height at the surface of the cantilever $y = \frac{t}{2}$ of the strain gauge sensor is at $y = \frac{t}{2} + ts$, where $t$ is the thickness of the beam and $ts$ is the thickness of the strain gauge sensor in mm. The change in length $L$ can be calculated as follows:
\[ dL = (r + y) d\phi = rd\phi + yd\phi = dx + yd\phi \] (3.2)

\[ dx(y) = (r + y) d\phi + rd\phi + yd\phi = dx(0) + y \frac{dx(0)}{r} \] (3.3)

where \( dx(0) \) is the natural section of the beam. The deformation can then be computed as:

\[ dx(y) - dx(0) = y \frac{dx(0)}{r} \] (3.4)

Hence the strain as a function of the height is:

\[ \varepsilon_x(y) = \frac{dx(y) - dx(0)}{dx(0)} = \frac{y}{r} \] (3.5)

The strain at the surface of the cantilever is:

\[ \varepsilon_x \left( \frac{t}{2} \right) = \frac{t}{2r} \] (3.6)

The strain at the surface of the sensor is:

\[ \varepsilon_x \left( \frac{t}{2} + ts \right) = \frac{t + ts}{r} \] (3.7)

The radius of curvature is related directly to the bending momentum \( M(x) \) as:

\[ \kappa = \frac{1}{r(x)} = \frac{-M(x)}{EI} \] (3.8)

where \( E \) is Young’s modulus of elasticity in \( Pa \) and \( M(x) \) is:

\[ M(x) = -F (L - x) \] (3.9)

\( I \) is the moment of inertia of the cross section area \( A \), in \( mm^4 \), as shown:

\[ I = \int y^2 dA = \frac{Wt^3}{12} \] (3.10)

The Von Mises equivalent stress \( \sigma' \) is equal to the normal stress \( \sigma_x \). The bending stress is given by
\[
\sigma_x = \frac{M(x)t}{2I} \quad (3.11)
\]

Therefore, the stress \(\sigma_x\) can be defined as

\[
\sigma_x = \frac{6F(x - L)}{Wt^2} \quad (3.12)
\]

The deflection of the beam \(\delta\) in \(mm\) can be derived from Euler-Bernoulli relationship and is given by

\[
\delta = \frac{Fx^2(x - 3L)}{6EI} \quad (3.13)
\]

The maximum deflection occurs at the point load \(F\) and is given by

\[
\delta_{\text{max}} = \frac{FL^3}{3EI} \quad (3.14)
\]

So, the force can be written as:

\[
F = \frac{3E1\delta_{\text{max}}}{L^3} \quad (3.15)
\]

The strain \(\varepsilon_x\) at the surface of the cantilever is:

\[
\varepsilon_x\left(\frac{t}{2}\right) = \frac{3t\delta_{\text{max}}(L - x)}{2L^3} \quad (3.16)
\]

The strain \(\varepsilon_x\) at the surface of the strain gauge is:

\[
\varepsilon_x\left(\frac{t}{2} + ts\right) = \frac{3(\frac{t}{2} + ts)\delta_{\text{max}}(L - x)}{L^3} \quad (3.17)
\]

As mentioned earlier, the dimensions of the beam cantilever were predetermined by the dimensions of the tensilometer as shown in Table 3.1.
Table 3.1: Aluminium beam dimensions.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Length $L$</td>
<td>80</td>
</tr>
<tr>
<td>Width $W$</td>
<td>40</td>
</tr>
<tr>
<td>Thickness $t$</td>
<td>1</td>
</tr>
<tr>
<td>Stress location $x$</td>
<td>10</td>
</tr>
</tbody>
</table>

The maximum displacement (deflection) allowed by the tensilometer is $5\text{mm}$. Hence, calculating the strain at the surface of the beam yields:

$$\varepsilon_x \left( \frac{t}{2} \right) = \frac{3(1\text{mm})(5\text{mm})(80\text{mm} - 10\text{mm})}{2(80\text{mm})^3} = 0.00102$$  \hspace{1cm} (3.18)

Calculating the strain at the surface of the sensor yields:

$$\varepsilon_x \left( \frac{t}{2} \right) = \frac{3\left( \frac{1\text{mm}}{2} + 100\text{um} \right)(5\text{mm})(80\text{mm} - 10\text{mm})}{(80\text{mm})^3} = 0.00123$$  \hspace{1cm} (3.19)

3.3.2 Finite Element Analysis of Beam Deflection

COMSOL Multiphysics® is a finite element analysis (FEA) software used to simulate and visualize the beam with the dimensions as mentioned in Table 3.1. The simulation helps visualizing the stress on the surface of the beam in order to determine the optimal location for the strain gauges to be placed. Fig. 3.5 shows the planar stress on the $xy$-plane of the beam when deflected by $5\text{mm}$. 

35
Figure 3.5: FEA of a deflected beam with dimensions shown in Table 3.1.

Fig. 3.6 shows a graph of the stress along the beam when deflected by 5mm. This graph can be used to approximate the surface stress acting on the strain gauges. Note that the beam in the simulation is clamped at 80mm, meaning that the ending at 0mm is free to move.

Figure 3.6: Stress along the beam as a function of the length of the beam
Chapter 4

Results and Discussion

This chapter presents the experimental results of the designed structures. First, tensile testing of the substrates results are shown to determine the elastic modulus of Kapton and VHB. A brief comparison of the fabrication technologies are presented with topography measurements. Then, results of the piezoresistivity characterization of PEDOT:PSS and silver strain gauges are presented. Finally, a comparison of the results with a commercially purchased strain gauge are presented with a discussion of the results.

4.1 Substrate Material Comparison

In order to determine the mechanical properties of the substrates used for the strain gauges, it is important to determine the elastic modulus. Tensile testing using Bose Bi axial ElectroForce is performed as shown in Fig. 4.1.
Figure 4.1: (a) Tensile testing using the Bose ElectroForce tensilometer, and (b) a closer look showing the clamped substrate.

The purpose of this test is to determine the level of hysteresis in the chosen substrates. Each of the substrates is clamped in the machine while the force is controlled and the displacement is measured simultaneously. The test was done three times for each substrate in order to see any variations of the results in between the tests. Fig. 4.2 shows the force applied on the 2 layers of Kapton and Fig. 4.3 shows the change in displacement due to the force applied in one of the samples.

Figure 4.2: Controlled force applied to 2 layers of Kapton.
Figure 4.3: The typical displacement measured simultaneously while the force in applied.

The thickness of the Kapton tape used as the substrate is 2.5um and the width of the tape is 1.27cm. Knowing the thickness $t$ and the width $w$ of the substrate we can calculate the stress applied as follows

$$\sigma = \frac{F}{A} = \frac{F}{wt}$$

where $F$ is the force applied on the substrate $A$ is the area. The stress $\sigma$ can now be plotted as a function of strain $\varepsilon$ to extract the tensile modulus of the material. Fig. 4.4 shows the best fit stress vs strain curve of the double layer Kapton substrate.
Figure 4.4: Best fit stress vs strain curve of a double layer Kapton substrate.

The slope of the graph is the elastic modulus of the material under test and is calculated to be $E = 1.3 \pm 0.08\,\text{GPa}$.

Similarly, tensile testing was done on the VHB substrate. Hysteresis stress vs strain curve of VHB is shown in Fig. 4.5. The curve clearly shows a lot of hysteresis in the elastoviscous material. The time required for the substrate to return to its original length is not calculated in this work but reported elsewhere to be up to 141s [22]
We can first notice that VHB requires much less force in order to produce the same displacement as Kapton, this is due to the flexibility of the elastomer. However, we see from the VHB figures that hysteresis is clearly presented.

For this reason, Kapton was considered as the main substrate material used for the design of the strain gauges. However, VHB was used later to mount the sensors on the skin for a biomechanics application of the designed strain gauges.

4.2 Comparison of the Fabrication Technologies

As mentioned earlier, two fabrication techniques were used to fabricate the PEDOT:PSS strain gauges—namely aerosol jet printing and laser micromachining. The two technologies are compared in terms of the thickness of the sensors produced and resulting electrical resistances of the patterned structures. While the aerosol jet printed can achieve printing resolutions down to 10\(\mu\)m in width, the printing resolution depends on the material used, the atomization method, the substrate and the width of the nozzle. The smallest thickness of PEDOT:PSS lines achieved using the aerosol jet printed was approximately 70\(\mu\)m, due to the viscosity of the material. It is worth noting that the PEDOT:PSS was diluted in DI water with 1:1 ratio prior ultrasonic atomization. The nozzle used to deposit the PEDOT:PSS lines has a diameter of 200\(\mu\)m. One of the main problems of diluting...
PEDOT:PSS in water and depositing the lines using the aerosol jet printer is once the lines dry, some areas in the line does not have PEDOT:PSS content in them-causing the line to be non-conductive as shown in Fig. 4.6.

![Image](image_url)

**Figure 4.6:** A dry printed PEDOT:PSS line using aerosol jet printer with some missing areas.

For this reason and also to increase the thickness of the lines, in order to reduce the initial resistance, multiple printing runs were required to achieve thicker lines. Fig. 4.7 shows the topography of a typical printed line on Kapton as a substrate using aerosol jet printing taken using the white light interferometry module of Polytec MSA-500 measurement equipment.
As shown in Fig. 4.7 (a), the thickness of the line, after 3 runs at a deposition speed of $1 \frac{\text{mm}}{s}$, is approximately 1 $\mu$m. Another issue of printing PEDOT:PSS using the aerosol jet printer is the temperature effect on the ink. The PEDOT:PSS ink is a solution dried inside the carrier tube during idle time, which in effect increased the tube pressure. To avoid this problem, the printing of the ink had to be done fast before the PEDOT:PSS start drying inside the tube. To avoid this issue, purging the tube after every printed line is required.

Laser micromachining is another way used to fabricate both PEDOT:PSS and silver paint strain gauges. The minimum reported resolution of the laser used is 20 $\mu$m. However, in order to reduce the resistance of the structure, the width and the thickness of the ink were increased by printing more lines on top of the existing lines. Again, Polytec white light interferometry was used to acquire topography measurements of both PEDOT:PSS and silver paint strain gauges as shown in Fig. 4.8 and Fig. 4.9, respectively.
Figure 4.8: (a) A 2-dimension topography picture showing the thickness of the PEDOT:PSS laser micromachined structure and (b) a 3-dimension look into the same structure.
Figure 4.9: (a) A 2-dimension topography picture showing the thickness of the silver paint structure and (b) a 3-dimension look into the same structure.

From the topography measurements in Fig. 4.8, the uniformity of the PEDOT:PSS thickness is shown with an average thickness value of approximately 40 $\mu m$. The measured thickness of the silver paint structure is approximately 35 $\mu m$ as shown in Fig. 4.9.

The initial resistance of each of the sensors built with different fabrication techniques is measured and shown in Table 4.1. Note that due to the variability of the PEDOT content in the PEDOT:PSS solution and the deposition method, the initial resistance value varies significantly.

Table 4.1: Initial structure resistance

<table>
<thead>
<tr>
<th>Fabrication method</th>
<th>Value [Ω]</th>
<th>Material</th>
<th>Thickness [μm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Aerosol jet printing</td>
<td>149000</td>
<td>PEDOT:PSS</td>
<td>1 ± 0.2</td>
</tr>
<tr>
<td>Laser etching</td>
<td>344 ± 130</td>
<td>PEDOT:PSS</td>
<td>40 ± 5</td>
</tr>
<tr>
<td>Laser etching</td>
<td>102 ± 10</td>
<td>Silver paint</td>
<td>35 ± 5</td>
</tr>
</tbody>
</table>
4.3 Electromechanical Characterization of PEDOT:PSS Strain Gauges

A PEDOT:PSS strain gauge was mounted on an aluminium beam in order to characterize the electromechanical characteristics of the sensor. As seen from the FEA model of the beam in Section 3.3.2 Fig. 3.6, the highest stress on the surface occurs near the fixed end of the beam. Hence, the sensor is mounted on the area with highest stress. The area of interest in the beam was initially rubbed using isopropanol alcohol to clean the surface in preparation for the sensor mounting. Then, a small amount of super glue adhesion is applied to the back of the sensor and laid on the aluminium beam surface. Once laid on the surface, a small finger pressure is applied on the sensor for 5 minutes to ensure good adhesion between the surface and the sensor. Finally, light copper wires were soldered on the copper tape pads of the sensor.

After the sensor was mounted on the beam, the tensile testing is performed by controlling the displacement, while measuring both the force and the voltage. The maximum displacement allowed by the tensilometer is 5mm. Consequently, sinusoidal waves of varying frequencies with an amplitude of $0 - 5\text{mm pk} - \text{pk}$ were applied to the aluminium beam, as shown in Figure 4.10. While many different actuating frequencies were attempted, two actuating frequencies of $0.1\text{Hz}$ and $0.2\text{Hz}$ were chosen, as they show the most predictable behaviours.
As mentioned earlier, 4-point measurement technique was adapted to avoid contact resistances in the measurements. Current of values ranging from $1 - 10mA$, depending on the structure’s resistance, are applied to the structure. The voltage read in the tensilometer using a 16-bit analog to digital (ADC) converter is amplified by the following equation, as given in the tensilometer manual

$$Gain = \frac{6k}{R_{\text{chosen}}} + 1$$  \hspace{1cm} (4.2)

The value of the resistance was chosen to be $R_{\text{chosen}} = 2k\Omega$ for a voltage gain of $gain = 4$. The force applied on the aluminium beam along with the voltage response to the change in displacement for the two actuating frequencies of $0.1Hz$ and $0.2Hz$ are shown in Figure 4.11 (a), and (b) respectively.

**Figure 4.10:** Displacement actuation of the aluminium beam at a frequency of 0.1 Hz.
Figure 4.11: The force and voltage responses to a deflection of 5 mm at (a) 0.1 Hz and (b) 0.2 Hz.

Note that the response of the sensor was not seen in the compression state, when the load is negative. Therefore, only the positive part of the signal was taken into account for curve fitting purposes. As can be seen from the figures, the voltage clearly has the same period as the applied force period. From the variation of the voltage, we can extract the gauge longitudinal gauge factor of the sensor. From the displacement and load results, we calculate the Young modulus of the beam with
the strain gauges mounted on it, as follows:

$$E = \frac{FL^3}{3wt^3 \delta_{\text{max}}} = \frac{(6.1N)(80\text{mm})^3}{3(40\text{mm})(1\text{mm})^3(2.5\text{mm})} = 135.1\text{GPa} \quad (4.3)$$

The stress absorbed by the surface on which the strain gauge is mounted

$$|\sigma_x| = \frac{6F(x - L)}{Wt^2} = \frac{6 \times (13.2N)(10\text{mm} - 80\text{mm})}{(40\text{mm})(1.1\text{mm})^2} = 114.55\text{MPa} \quad (4.4)$$

Hence, the strain on the surface is the deviation of the stress by the tensile modulus

$$\varepsilon_x = \frac{|\sigma_x|}{E} = \frac{114.55\text{MPa}}{135.1\text{GPa}} = 0.00085 \quad (4.5)$$

As mentioned earlier, the longitudinal gauge factor $G_{||}$ can be calculated as follows

$$G_{||} = \frac{\Delta V}{V \varepsilon_x} = \frac{4.299V - 4.293V}{4.293V \frac{1}{0.00085}} = 1.6 \quad (4.6)$$

Now that we extracted the longitudinal gauge factors $G_{||}$, we can extract the piezoresistive coefficient as follows

$$G_{||} = (1 + 2v + \pi_{||}E_{\text{substrate}}) \quad (4.7)$$

where $v$ is the poisson ratio of the substrate (Kapton) and is given from the materials information sheet to be 0.33, $\pi_{||}$ is the longitudinal piezoresistive coefficient, and $E_{\text{substrate}}$ is measured in Section 4.1 to be 1.3GPa. Hence,

$$\pi_{||} = \frac{G_{||} - 1 - 2v}{E_{\text{substrate}}} = \frac{1.6 - 1 - 2(0.33)}{1.3 \times 10^9} = -4.6 \times 10^{-11}\text{Pa}^{-1} \quad (4.8)$$

The change in the resistance when the sensor is mounted transversally is very small. The transversal gauge factor $G_{\perp}$ and piezoresistive coefficient $\pi_{\perp}$ respectively can be calculated as

$$G_{\perp} = \frac{\Delta V}{V \varepsilon_x} = \frac{4.29V - 4.293V}{4.293V \frac{1}{0.00085}} = -0.822 \quad (4.9)$$

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The change in voltage is very small. In fact, it is within the noise level of the signal so the calculated value will be neglected.

4.4 Commercial Strain Gauges

The results of the PEDOT:PSS based strain gauges are compared to a commercial gauge. The force and voltage response of the commercial strain gauge is shown in Figure 4.13.

![Force and Voltage Response](image)

**Figure 4.12:** The force and voltage responses to a deflection of 5 mm at 0.1 Hz.

As can be seen from the figure, the response of the commercial strain gauge seems to be cleaner. However, the longitudinal gauge factor of the PEDOT:PSS of 1.6 is comparable to the calculated gauge factor of the commercial strain gauge of 1.84. For this reason, it was thought that a metallic ink can be incorporated in a design of a strain gauge using the same fabrication method as fabrication of the PEDOT:PSS strain gauges.
4.5 Electromechanical Characterization of Silver Strain Gauges

Same approach of mounting the silver paint based strain gauge was taken in order to characterize its piezoresistive properties. Displacement actuations at frequencies of 0.05Hz, 0.1Hz and 0.5Hz were applied while measuring the force and voltage. The longitudinal response force and voltage at the three frequencies is represented in Figure 4.14 a, b and c, respectively.

![Figure 4.14 a, b and c](image-url)
Figure 4.13: The longitudinal force and voltage responses of the silver paint strain gauge to a deflection of 5mm at (a) 0.05 Hz (b) 0.1 Hz and (c) 0.2 Hz.

The longitudinal gauge factor $G_{\text{silver}||}$ and the piezoresistive coefficient $\pi_{\text{silver}||}$ of the sensor is calculated to be

$$G_{\text{silver}||} = \frac{\Delta V}{V} \frac{1}{\varepsilon_x} = \frac{2.82V - 2.79V}{2.79V} \frac{1}{0.0085} = 12.6 \quad (4.10)$$

$$\pi_{\text{silver}||} = \frac{G_{\text{silver}||} - 1 - 2v}{E_{\text{substrate}}} = \frac{12.6 - 1 - 2(0.33)}{1.3 \times 10^9} = 8.4 \times 10^{-9} \text{ Pa}^{-1} \quad (4.11)$$

The transversal gauge factor $G_{\text{silver}\perp}$ and the piezoresistive coefficient $\pi_{\text{silver}\perp}$ of the sensor is calculated to be

$$G_{\text{silver}\perp} = \frac{\Delta V}{V} \frac{1}{\varepsilon_x} = \frac{2.775V - 2.772V}{2.772V} \frac{1}{0.0085} = 0.7220 \quad (4.12)$$

$$\pi_{\text{silver}\perp} = \frac{G_{\text{silver}\perp} - 1 - 2v}{E_{\text{substrate}}} = \frac{0.7220 - 1 - 2(0.33)}{1.3 \times 10^9} = -7.215 \times 10^{-10} \text{ Pa}^{-1} \quad (4.13)$$

Again, the change of voltage (for constant current) when the sensor is mounted
laterally is very small. Therefore, the calculated value will be neglected.

4.6 Comparison

The results of both the PEDOT:PSS and silver paint based strain gauges are compared to a commercial gauge. The longitudinal gauge factor of the PEDOT:PSS of 1.6 is comparable to the calculated gauge factor of the commercial strain gauge of 1.84. It should be noted that the gauge factor value provided by the manufacturer of the commercial strain gauge is 2.1. The resulted experimental gauge factor, on the other hand, could have possibly suffered from the stress not being fully transformed from the aluminium beam surface to the strain gauge itself. Since the mounting procedure of all the strain gauges is the same, the assumption of small loss in the stress from transforming completely to all the strain gauges should be valid for all. The silver paint strain gauge has experienced a much higher gauge factor of 12.6 in comparison to both the PEDOT:PSS and the commercial strain gauges, but the non-smooth variation in the electrical resistance seems to indicate reversible disconnections of the conductive paths under larger stress.

Table 4.2: Summary of Experimental Longitudinal Gauge Factor of PEDOT:PSS, Silver Paint and Commercial Strain Gauges

<table>
<thead>
<tr>
<th>Strain Gauge</th>
<th>Longitudinal Gauge Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>PEDOT:PSS</td>
<td>1.6</td>
</tr>
<tr>
<td>Silver Paint</td>
<td>12.6</td>
</tr>
<tr>
<td>Commercial</td>
<td>1.84</td>
</tr>
</tbody>
</table>

It should be noted that due to the design of the strain gauges fabricated in the lab, the length of the strain gauge is maximized to increase the sensitivity of the sensors in the longitudinal direction, while the transitional areas are small to eliminate the transversal loading effect on the sensors. Hence, both of the transversal gauge factors of the the PEDOT:PSS and silver paint based strain gauges shown on the table are very small. One of the main issues of the silver paint strain gauge, however, is the very small range of operation. Looking back at the resulted waveform of the silver paint strain gauges, we see high spikes at higher values of stress.
This is due to reversible cracks in the silver lines, causing a sudden increase in resistance for larger stress, thus limiting the operating range. In order to avoid this problem, the silver based strain gauge can only operate at a smaller range.

While the flexibility of PEDOT:PSS is considered a plus, the variability in the gauge factor results from one sample to another hinders the viability of the use of this polymer in sensing applications. The deposition method used in designing the PEDOT:PSS strain gauges may have caused random alignment of the PEDOT:PSS chains in the solvent. Therefore, it could be the reason in the high variability in the results of the PEDOT:PSS strain gauges. It should be noted that while the nominal initial resistances of each PEDOT:PSS structure varied greatly (in the range of 200 ohm to 10 kohm), the response varied between one sample to another. Some samples did not respond to the strain at all, but rather the resistance stayed constant. The data did not vary in time for the same PEDOT:PSS sample for repeated tests on the same day. The structures were not retested at different days to see if there is a variations in the results between day to day. The data shown in this thesis is of one structure that responded in this manner. The silver ink strain gauges results, however, did not have high variability between samples, proving the capability of the laser itching use in the patterning of conductive inks.

Another factor of comparison is the price of production. While the commercially purchased strain gauge costed $12 per gauge, the in-house fabrication of either the PEDOT:PSS or the silver point strain gauges costs approximately $3 per gauge. It should be noted that the price estimation of the in-house strain gauges is based on the cost of the materials purchased in small quantities. If, however, mass production is required, in bulk purchasing of the materials can potentially reduce the price down to $1 per gauge. Besides, the in-house fabrication time of the strain gauges can take up to a day per ~10 structures. It should be noted that the fabrication time depends on the curing time of the material. As a result, fabrication the silver paint strain gauges is actually faster than the PEDOT:PSS strain gauges.

Lastly, one of the main advantages of exploiting the alternative microfabrication techniques in the design of the strain gauges as discussed in the this thesis is the customizability of the design for specific applications. In order to utilize the strain gauges to their maximum sensitivity, variable strain gauges shapes maybe required. For example, strain sensing of human skins due to movements can bene-
fited from specified gauge shape designs to promote highest sensitivity.
Chapter 5

Conclusions

5.1 Thesis Overview

The main motivation of this work was to develop custom designed structures (strain gauges taken as target example) using alternative microfabrication techniques, based on either conductive polymers (PEDOT:PSS) or metallic inks such as silver. The goal for developing such sensors is to incorporate them in applications ranging from joint monitoring in humans to integration of such sensors in fabrics or sporting attire.

Initially, the idea of developing printable sensors was initiated by the acquisition of new pieces of equipment in Dr. Edmond Cretu’s Adaptive MEMS laboratory. It is reported in the literature in [37] [28] [20] [21] [23] [39] for PEDOT:PSS to have intrinsic piezoresistive properties. However, the reported gauge factors in the literature vastly varied from one paper to another. In addition, to the author’s knowledge, there is no attempts to characterize the transversal gauge factor of PEDOT:PSS. In order to claim intrinsic piezoresistivity of a material, it is thought that it is important to characterize both the longitudinal and transversal force response. Also, the fabrication techniques used in the literature are mostly conventional clean room processes, meaning a non-rapid slow manufacturing of sensors to be studied.

The first attempt in this work to design PEDOT:PSS strain gauges was done by using the Sonoplot Microplotter. The principle of operation relies on dipping a small glass tip in the solution, and due to capillary forces, the fluid fills the tube.
Then, through ultrasonic vibration in a piezoelectric element mounted on the glass tube, controlled amount of the solution is deposited on the material. Less success was seen depositing PEDOT:PSS using the microplotter due to several reasons, including: the viscosity of the material, the non-uniformity of the surface, the PEDOT:PSS content of the solution deposited. The deposition of PEDOT:PSS lines was then successful using the aerosol jet printer. The use of the aerosol jet printer proved to be a very good alternative to deposit PEDOT:PSS lines, but due to the very small thickness of the deposited ink, multiple deposition if required to achieve a thicker structure. The process of multiple depositions required close looks into the tube pressure of the equipment since the PEDOT:PSS dried in the tube, causing unexpected clogs. Finally, it was though that laser micromachining etching can provide the best results of patterning when it comes to strain sensors. Thick layers of ink is thought to be required for flexible printed strain sensors as it is important for the lines not to break when bent. Silver paint is then fabricated using the same method of laser micromachining etching as an alternative material.

Using laser micromachining, both PEDOT:PSS and silver paint strain gauges were patterned and characterized longitudinally and transversally using controllable tensile testing. Four-point measurement methodology was adapted in order to roll out contact resistances from the gauge factor measurements. Tensile testing was initially done on different substrates including Kapton and VHB. Due to the great hysteresis found in VHB, Kapton was chosen as the main substrate material for the strain gauges. The in-house designed sensors were mounted on an aluminium beam along with a commercially available strain gauge. While the displacement of the tensilometer was controlled, both the force applied on the aluminium beam and the voltages read from the strain gauge resistors were read simultaneously. The average resultant longitudinal and transversal gauge factors of the PEDOT:PSS strain gauges were 1.6 and -0.82, respectively, with a large variation from sample to sample. On the other hand, the longitudinal and transversal gauge factors of the silver paint sensors were 12.6 and 0.722, respectively. Although the PEDOT:PSS strain gauges have a small longitudinal gauge factor, they are very stretchable, making them suitable for applications requiring a large range of operation. In fact, it was reported in the literature a stretchability of PEDOT:PSS of 17%. One of the main issues, however, of the PEDOT:PSS strain gauges is the
inconsistency in the measurements. It is believed that is caused due to the align-
ment of the chains in the synthetic polymer. As a result, different fabrications and
depositions techniques result in inconsistent response of the sensor. Nonetheless,
the work in this thesis has shown the potential for using alternative deposition and
patterning methods to fabricate very cheap disposable sensors.

The main contribution of this work was to establish methods of patterning con-
ductive inks on flexible substrates. The deposition of PEDOT:PSS on the VHB
elastomer is one of the few examples of the capabilities of the established alterna-
tive fabrication technologies investigated in this thesis. The use of sensors designed
using the studied alternative fabrication technologies can be used in biomedical
applications. An example application will be relating EMG to force measurements
by mounting a custom patterned force sensor on the skin of a person and seeing the
effects of flexing the muscle while monitoring EMG.

5.2 Future Outlook

In this work, a great deal of attention on the investigation of the piezoresistivity
of PEDOT:PSS was given while exploring different fabrication methods to deposit
different kinds of inks on flexible substrates. While the thorough investigation of
PEDOT:PSS as a synthetic material to be used in strain gauges proved so far that
it might not be a suitable material in terms of both sensitivity and reproducibility
of the structures, the fabrication methods investigated showed the potential to be
used with other types of inks. Mixing the synthetic polymer with other organic
materials and using the same fabrication methods can seems the next logical step.
While PEDOT:PSS in fact seems to be affected only by geometrical changes, the
conductivity, flexibility, and the ease of processability of the material are just a few
advantages. It is reported elsewhere that a mixture of PEDOT:PSS with Polyvinyl
alcohol (PVA) increases the stiffness of the material and results in a gauge factor
as high as 396 [16] [26].

Another alternative to intrinsically piezoresistive sensors, the geometrical ef-
fect can be incorporated in designing a capacitive sensing strain sensor. The idea
is to pattern small capacitive combs on flexible material. The change of the over-
lapping area of the capacitor combs is a result strain. A study published in 2014
showed the potential for developing capacitive metallic strain gauges for wireless monitoring [46]. Power consumption is the main advantage of developing such sensors, as operating principle relies on a sensing capacitance dependence on the applied strain. While the published study uses metallic capacitive combs, the use of a synthetic polymer to design a capacitive strain gauge can prove to be a cheaper option.

Further research is also required for musculoskeletal characterization. While EMG monitoring has always proved to be a method to monitor a muscle group activity, EMG acts only as an input to the system. Hence, EMG alone does not give the full picture about the force produced in the muscle. Strain measurements along with EMG can help in identifying the system. Further research is required to first identify most sensitive strained areas, for a strain sensor to be mounted, in response to a specific muscle group activation. Also, since strain measurements are conducted on the skin of the subject, it is essential to monitor all the muscle groups around the area of interest.
Bibliography


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Appendix A

Supporting Materials

In this chapter, a potential application of designed strain gauges is presented. The PEDOT:PSS strain gauge, due to its flexibility, is chosen to monitor the deflection in the skin around the bicep brachii, while integrated EMG is read off the muscle group. The application presents the potential of dealing with the muscles response as a system. That is, the input to the muscle is the EMG and the output in this case is deflection. Also, this implementation of a strain gauge sensor on the body presents the possibility to integrate custom designed sensors for many human monitoring applications including but not limited to: sensor integration in fabrics, and body area networks. In this chapter, a brief explanation of the implementation of National Instruments (NI) LabView Virtual Instrument (VI) on a Field programmable gate array (FPGA) is described. Then, the experimental setup of measuring the integrated EMG signal along with the strain measurements is presented. Finally, the measurement results are presented along with a discussion.

A.1 LabView Virtual Instrument (VI) Design

The NI LabView FPGA allows for a simple and rapid system level implementation for verification of a system. The LabView FPGA module implements a graphical programming technique that can be transformed into a low-level of abstraction through the LabView Embedded technology. The program design is implemented using an FPGA module along with NI reconfigurable I/O (RIO) hardware. Due to
the high parallelism and flexibility LabView provides for FPGA programming, it is thought that using such system is ideal for this experiment.

The system designed in LabView encompasses two Virtual instruments (VIs), namely: a host and a target. While the host encompasses all the signal processing and the graphical user interface (GUI), the target acts as an adaptive data acquisition unit implemented on an FPGA. Fig. A.1 (a) shows the GUI and (b) the VI implementation of the host and Fig. A.2 shows the implementation of the target VI.
Figure A.1: (a) The graphical user interface and (b) the implementation of the LabView Host VI
The communication between the host and the target VIs is done through a direct memory access (DMA) first-in-first-out (FIFO). This allows the target VI to use the host RAM directly, allowing for a major improvement in the speed of acquisition. The FIFO used has two parts, the first of which is in the target and is used to save the data read through the IO channel, and the second is in the host, which is used to read out the data when the FIFO is full. The sampling rate of data acquisition is specified by the loop rate of the target VI. For this experiment, it is specified at 100µsec. The LabView target VI is also configured to output two analog signals specified by the user in the range of ±10V to power up and external devices.

**Figure A.2:** The implementation of the LabView Target VI
A.2 Experimental Setup

The experimental setup consists of an integrated EMG amplifier (Advancer Technologies Muscle Sensor v3) from Pololu Robotics and Electronics©, general-purpose, disposable Ag-AgCl EMG surface electrodes, a PEDOT:PSS strain gauge, current source, and VHB elastomer for mounting purposes. First, the skin was prepared for electrode placements using alcohol pads and abrasion using Nuprep gel. This step is very important in order to provide better skin adhesion and to reduce the skin electrical resistance. Two differential electrodes are then placed on the bicep brachii muscle and the third electrode is placed on the elbow bone as a reference. Once the electrodes are placed on the skin, a small piece of VHB tape is used to mount the PEDOT:PSS strain gauge on the skin. VHB was chosen due to its high flexibility and adhesion to the skin. The configuration of the electrodes and the strain gauge are shown in Fig. A.3. In the same figure we see the flexing of the relaxation state and the flexing state of the bicep muscle.
Figure A.3: The strain gauge mounted on the skin of the bicep in (a) relaxed state, and (b) flexed state.

The EMG amplifier requires powering signals of $\pm 5\, \text{V}$, provided by the LabView system and current is applied through the strain gauge resistor while monitoring the voltage by the LabView VI.

### A.3 Measurement Results

With a similar setup, another research team have used a silicon based strain gauge to develop a sensor, called muscle contraction (MC) sensor, to measure muscles contractions, as mentioned in [47]; their results are shown in Figure A.4.
**Figure A.4:** Simultaneous recording of the force (Fg), MC and EMG. The Fg and MC variables are normalised to the maximal value, adapted from [47]

As shown in Fig. A.5, our measurement results of the induced strain (using PE-DOT:PSS strain gauges) indicate a certain level of correlation with the concurrent EMG measurements.
Figure A.5: The results of contraction (a) strain data, and (b) integrated EMG.

One of the main challenges in performing such correlation is to find the perfect area of the skin that experiences the most stress due to flexing the bicep brachii. Another main challenge is establishing the right functional dependence between the skin stretching and the activation of the muscles. In this experiment, it was noticed that movement of the skin due to the tricep muscle activation caused some spikes in the strain measurements. However, focusing on the results between 30 sec and 50 sec in Fig. A.4, we notice a clear correlation between the EMG of the bicep brachii and the strain. Having other spikes in the results should not be surprising,
as the deflection of the skin (i.e. the stress applied on the strain gauge mounted on the skin) is a function of not only the input EMG to the bicep muscle, but rather a function of multiple muscles inputs.

In order to find the cross correlation between the input signal EMG and the output signal (strain), we use the Mathworks® MATLAB command \texttt{xcorr}. The correlation coefficients represent the matching between the input signal while shifted across the output with highest correlation found at zero delay of approximately 60%.
Appendix B

Publications

Although not directly related to my thesis work, some other activities during my masters resulted in 2 conference publications:

1- Guillén-Torres, M. Á., Almarghalani, M., Sarraf, E. H., Caverley, M., Jaeger, N. A., Cretu, E., & Chrostowski, L. (2014, September). Silicon photonics characterization platform for gyroscopic devices. In Photonics North 2014 (pp. 92880U-92880U). International Society for Optics and Photonics. I created the hardware, firmware and software interfaces of the rotation table, as well as coding the rotation patterns. I wrote part of the manuscript, commented and edited the full final manuscript.