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Flexible Capacitive Pressure Sensors and Triboelectric Energy Harvesters Using Laser-Assisted Patterning Process for Flexible Hybrid Electronic Applications

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FLEXIBLE CAPACITIVE PRESSURE SENSORS AND TRIBOELECTRIC ENERGY HARVESTERS USING LASER-ASSISTED PATTERNING PROCESS FOR FLEXIBLE HYBRID ELECTRONIC APPLICATIONS

by

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A thesis submitted to the Graduate College in partial fulfillment of the requirements for the degree of Master of Science in Engineering Electrical and Computer Engineering Western Michigan University June 2020

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FLEXIBLE CAPACITIVE PRESSURE SENSORS AND TRIBOELECTRIC ENERGY HARVESTERS USING LASER-ASSISTED PATTERNING PROCESS FOR FLEXIBLE HYBRID ELECTRONIC APPLICATIONS

Valliammai Palaniappan, M.S.E.
Western Michigan University, 2020

This work focuses on the design, fabrication and characterization of novel flexible capacitive pressure sensors and triboelectric energy harvesters using laser-assisted patterning process for flexible hybrid electronic applications. Initially, the capacitive pressure sensor was developed by fabricating a set of polydimethylsiloxanes (PDMS) dielectric films with pyramid shaped micro-structures using a laser-assisted patterning process. The pressure sensor consists of two electrodes (top and bottom) that were fabricated by depositing silver (Ag) on flexible polyethylene terephthalate (PET) using additive screen-printing process. Finally, the pressure sensor was assembled by attaching the top and bottom Ag electrodes to the smooth side of pyramid shaped micro-structured PDMS dielectric layers. The capability of the fabricated pressure sensor was investigated by subjecting the sensor to pressures ranging from 0 to 10 kPa. In addition, the stability of the fabricated pressure sensor towards the applied pressure, response and recovery time of the pressure sensor, as well as the hysteresis of the fabricated pressure sensor was also studied.

Then, a flexible vertical contact-separation mode based triboelectric nanogenerator (TENG) was fabricated using laser-assisted patterning and additive screen-printing processes. A micro-pyramid patterned polydimethyl siloxane (MP-PDMS) was prepared by using a laser engraved acrylic mold. A top and bottom electrode for TENG was fabricated by screen printing silver (Ag) on a flexible polyimide substrate. The MP-PDMS, was bonded to the bottom electrode using a thin layer of adhesive PDMS. The MP-PDMS and the printed polyimide electrodes were employed as positive and negative triboelectric layers, respectively to scavenge mechanical energy into electrical energy. The performance of the MP-TENG was demonstrated by investigating the open-circuit voltage, short-circuit current and instantaneous power on the load resistor.
ACKNOWLEDGMENTS

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Valliammai Palaniappan
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CHAPTER I
INTRODUCTION

1.1. Background

Ultra-thin, conformal and physically flexible electronic products have been under development for years, and the recent advances in materials and methods have improved the capability to manufacture flexible hybrid electronic (FHE) systems [1,2]. FHE is the collaboration of two industries: the electronics industry and the printing industry, which have provided a unique opportunity to design novel FHE devices for improving human lives through advanced human healthcare monitoring [3] and human machine interfacing [4] applications. Flexible pressure sensors play a key role in sensing physiological signals such as respiration rate, wrist and carotid pulse, and heart-beat rate [5-12]. Other applications of flexible pressure sensors include touch-screen sensing in smart-phone LED [13] and LCD displays [14], and microphones applications [15] to monitor the air flow by sensing pressure throughout the system in the commercial and residential buildings.

For FHE devices, the power sources also need to be designed and developed with soft, flexible, bendable and stretchable form factors to provide low-cost flexible power solution for next generation devices [16,17]. Energy harvesting techniques have been employed to harvest ambient energy from various environments such as wind, solar, waves/tides, hydro mass and biomass. [18-20]. Therefore, to summarize, the author has explored the development of wearable flexible pressure sensors for improving health monitoring standards and a flexible energy harvester for powering FHE devices.
1.2. Author’s Contribution

*Conference papers:*


Journal:


1.3. Organization of Thesis

The rest of this thesis is divided into five chapters. In Chapter II, the author presents a comprehensive literature review. This review provides the basic introduction to FHE and various fabrication techniques of FHE. This is followed by a background about wearable pressure sensing technology and energy harvesters. In Chapter 3 and 4, the author discusses in detail two research projects that have been performed for developing a flexible micro-structured capacitive pressure sensor and a flexible triboelectric energy harvester, respectively. Chapter 3 includes details about the design, fabrication steps, characterization, test set-up used, and results of a fully fabricated micro-structured and hybrid capacitive pressure sensor. Chapter 4 presents a discussion on a novel energy harvester - triboelectric nanogenerator (TENG). It includes details about the TENG fabrication, test set-up, and results. Finally, Chapter 5 summarizes this work with conclusions and suggestions for future research and development.
CHAPTER II
LITERATURE REVIEW

2.1. Introduction

This chapter is divided into three sections followed by a summary. Section 2.2 provides an introductory discussion about flexible hybrid electronics (FHE). This section provides a general overview of material aspects for FHE and the fabrication techniques for the development of FHE devices. Section 2.3 provides a review on wearable pressure sensors used for remote health monitoring followed by pressure sensing mechanisms; highlighting micro-structured capacitive pressure sensing. Section 2.4 provides a review on energy harvesting solutions for FHE devices followed by different types of energy harvesting mechanisms; highlighting the importance and need of triboelectric nanogenerator (TENG) for harvesting mechanical energy.

2.2. Flexible Hybrid Electronics (FHE)

FHE is an emerging technology with huge market potential that is estimated to grow up to ~$73.3 billion in 2029 [21]. FHE has gained considerable attention because the form factors of electronics have started to change from their traditional rigid and rectangular shapes to more soft, flexible, bendable and stretchable form factors [22,23]. Researchers are using the new form factors (soft, flexible, stretchable and bendable) of FHE to develop next generation electronics which are aimed to be flexible and conformal to the curves of a human body, and stretch across the shape of objects or structures, while preserving the full operational integrity of conventional electronics [24,25]. With this technology, there is a growing demand for sensors such as strain sensors [26-29], pressure sensors [30-32], humidity sensors [33-35], temperature sensors [36-38], and chemical sensors [39-41].
2.2.1. Material aspects of FHE

Flexible and stretchable substrates are the most important materials for FHE [42]. Most commonly reported FHE devices based on flexible/stretchable substrates utilize polymers, rubber, fiber and textiles [43]. Textile based fabrics, that utilize fibrous materials whose properties are determined by the properties of structure of the fiber or yarn of the fabric, are the most preferable choice of material for wearable applications [44-46]. Polymers and rubber substrates have also been intensively researched due to their thermal stability and chemical resistance [47,48]. Several research groups have fabricated FHE devices on flexible and stretchable polymers such as polyimide (PI), polyethylene terephthalate (PET), polydimethy siloxane (PDMS), polypropylene (PPy), poly(3,4-ethylenedioxythiophene) polystyrene sulfonate (PEDOT:PSS), and polyethylene naphthalate (PEN) [49]. Table 2.1 summarizes the chemical and physical properties of these
flexible and stretchable polymers and Figure 2.1 shows the molecular structures of the typically used flexible and stretchable polymers.

Table 2.1. Summary of physical and chemical properties of polymers used as substrates in FHE [49].

<table>
<thead>
<tr>
<th>Materials</th>
<th>Stretchable/bendable</th>
<th>Transparency/Dielectric constant</th>
<th>Thermal stability</th>
<th>Chemical resistance</th>
</tr>
</thead>
<tbody>
<tr>
<td>PI</td>
<td>Bendable</td>
<td>Low transparency 2.8-3.5</td>
<td>Resist temperature (&lt;450 °C)</td>
<td>Weak acids and alkali</td>
</tr>
<tr>
<td>PET</td>
<td>Bendable</td>
<td>High transparency 2.5-3.5</td>
<td>Resist temperature (&lt;100 °C)</td>
<td>Ethanol and acetone</td>
</tr>
<tr>
<td>PEN</td>
<td>Bendable</td>
<td>High transparency 2.9-3.2</td>
<td>Resist temperature (&lt;180 °C)</td>
<td>oxygen and water</td>
</tr>
<tr>
<td>Silicone (PDMS, Ecoflex)</td>
<td>Bendable and stretchable</td>
<td>High transparency 2.3-2.8</td>
<td>Resist temperature (&lt;100 °C)</td>
<td>Ethanol and acetone</td>
</tr>
</tbody>
</table>

2.2.2. Fabrication Techniques

The electronic industry is progressing towards FHE by employing additive manufacturing processes. The benefits of additive manufacturing allow for greater flexibility, when compared to traditional components, which enables printing electronic components on flexible substrates [50-53]. To fabricate FHE devices, there are five main fabrication processes that are typically used which include screen printing, gravure printing, flexography printing, inkjet printing, and laser etching [54].
2.2.2.1. Screen Printing

The process of screen-printing is shown in Fig. 2.2. The screen-printing process uses a mesh screen and printable functional ink to fabricate devices on flexible substrates. The substrate is aligned underneath the mesh screen and the ink is squeegeed across the mesh screen forcing the ink through the screen, thus creating the desired pattern or design on a substrate in a controlled and efficient manner. Screen-printing can deposit high viscosity inks (500-5000 cP). [55].

Figure 2.2. Screen-printing process

2.2.2.2. Inkjet Printing

Inkjet printing of electronics can be categorized into drop on demand inkjet printing and continuous bias inkjet printing. In continuous inkjet printing, continuous droplets of ink are
generated, which may be electronically controlled. Here, some of the ink droplets are subjected to electrostatic charge and are deflected to get a negative print by electrostatic deflectors while the uncharged droplets are printed to obtain the desired image onto the substrate. On the contrary, the drop-on-demand (DOD) inkjet printing system (Fig. 2.3) generates ink droplets only for the required image areas in order to obtain the desired print. This printer usually employs two techniques: thermal or piezoelectric. In a thermal inkjet printer, a vapor bubble is created due to the vaporization of the ink. This vapor bubble is used to push the ink out of the nozzle and onto the substrate. However, in piezoelectric inkjet printing system, ink droplets are pushed out when the piezoelectric material (printing head) undergoes mechanical deformation due to applied electric potential. This in turn changes the volume of the drop nozzle and the ink is pushed out [56].

2.2.2.3. Gravure Printing

Gravure printing is most promising printing process in terms of resolution and print speed. Figure 2.4 shows the gravure printing process. The gravure cylinder has small cells on the surface that are responsible for carrying ink from the ink fountain to the substrate. The gravure cylinder (image carrier) rotates in the ink fountain as a result of which cells get filled with ink. The doctor blade wipes off the excess ink that remains on the surface of the cylinder. Transferring the ink from the cells onto the substrate is assisted by the impression cylinder [57].
2.2.2.4. Flexography Printing Process

In this method, ink is transferred from an ink bath to a desired substrate when a fountain roller transfers the ink from ink bath to anilox roller. Then, a printing plate cylinder transfers controlled amounts of ink from the anilox roller to the substrate by controlling the pressure between the printing plate cylinder and an impression cylinder (Figure 2.5). This method is suitable for large...
scale manufacturing and to a variety of substrates, ranging from metal, plastics, and paper substrates [58].

2.2.2.5. Laser Etching

Laser etching is a process that uses a laser source to engrave or leave marks on an object (laser writing) by foaming, melting or ablation. This technique does not involve the use of any functional ink [59].

Typically, the laser etching process is used in industrial applications to create cell patterns in the flexographic plates, cylinders, and anilox rolls [60-62]. Conventional photolithography techniques have been extensively used to pattern and perform surface morphology modification of FHE devices, [63,64]. However, this is relatively time-consuming, involves complex steps and requires expensive equipment and clean-room facilities [65]. Therefore, laser etching based fabrication steps have been adopted for patterning FHE devices. Figure 2.6 shows the laser etching process:

![Laser etching process](image)

**Figure 2.6.** Laser etching process: (a) laser-marking, (b) laser cutting, (c) laser engraving
Laser etching typically involves laser marking, laser cutting, and laser engraving processes. Laser marking process is used to write/mark on an object (Fig. 2.6(a)), whereas, laser cutting, and laser engraving can change the surface morphology of a material by melting away undesired parts of the material/substrate (Fig. 2.6(b,c)). In general, laser etching is done by importing a CAD file with patterns to etch away the material and create desired patterns on the material. The laser head moves in x and y directions during laser etching process to create patterns.

2.3. Wearable Pressure Sensing in Real-Time Health Monitoring

2.3.1. Wearable Health Monitoring

The necessity to design and develop new strategies and technologies to provide remote healthcare monitoring has paved the way for wearable sensing technology [66-68]. Wearable sensing technology includes the soft, flexible, bendable and stretchable sensors that can be integrated into textile fiber, clothes, and/or directly attached to the human body [69-72]. These sensors are capable of monitoring physiological signals such as electrocardiogram (ECG),
electromyogram (EMG), heart rate (HR), body temperature, electrodermal activity (EDA), arterial oxygen saturation (SpO2), blood pressure (BP) and respiration rate through daily human activities such as walking, running, and exercising [73-76]. Wearable health monitoring allows people to monitor their health remotely in a convenient, portable and comfortable manner rather than in centralized healthcare services such as hospitals (Fig. 2.7).

2.3.2. Pressure Sensing Mechanisms

2.3.2.1. Piezoelectric mechanism

Piezoelectricity is instantaneously generated in specific materials when they are under strain and pressure [77]. The electrical charges are generated when external pressure is applied to the piezoelectric materials. The most commonly used piezoelectric materials are ZnO nanowires, and lead zirconate titanate (PZT) and are widely investigated for the development of flexible pressure sensors due to the high piezoelectric constants [78,79]. S. Chen et al proposed a wearable hierarchical self-power piezoelectric pressure sensor for cardiovascular monitoring of a human subject (Fig. 2.8) [80]. This work not only enables the wearable sensor to record tiny pulses from the artery and heart but also endows it with the ability to monitor wide blood pressure ranges.

Figure 2.8. Real-time cardiovascular monitoring of the fabricated piezoelectric pressure sensor [80]
2.3.2.2. Piezoresistive mechanism

Piezoresistive materials change resistance to the flow of current when they are compressed or strained. Piezoresistive pressure sensors are mainly formed by three components: the active materials, flexible substrates and conductive electrodes [81]. The active component consists of various conductive materials such as metal nanoparticles and nanowires [82], conductive polymers [83], carbon nanotubes [84], and graphene [85]. Y. Song et al proposed a highly compressible piezoresistive pressure sensor for health monitoring. In this work, a sponge like carbon nanotube based polydimethyl siloxane (CNT-PDMS) was used as the active sensing material. The relative resistance of sponge-like CNT-PDMS material changes due to its large deformation, for varying applied pressure. Figure 2.9 shows the highly compressible CNT-PDMS based piezoresistive pressure sensor that can detect or sense human physiological signals [86].

![Piezoresistive pressure sensor for physiological health monitoring](image)

Figure 2.9. Piezoresistive pressure sensor for physiological health monitoring [86]

2.3.2.3. Capacitive mechanism

A capacitive mechanism is one of the most important mechanism in sensing pressure and strain as it is reported to be highly sensitive, when compared to other mechanisms [87]. Typically, a dielectric elastomer material is sandwiched between two conductors. When pressure is applied. The increase in capacitance corresponds to the pressure sensitivity of the pressure sensor. S.
Sharma et al fabricated a highly sensitive, stable and flexible capacitive pressure sensor by sandwiching MaxenePoly (vinylidene fluoride-trifluoroethylene) (PVDF-TrFE) as a dielectric layer in between poly (3, 4-ethylenedioxythiophene) polystyrene sulfonate (PEDOT:PSS), the conductors move closer as the dielectric layer deforms, and the capacitance increases. This work demonstrated a high sensitivity of 0.51 kPa$^{-1}$, and wide pressure ranges up to 300 kPa. The fabricated sensor was attached to a wrist’s dermal area and the pulse rate recorded from the subject was 82 beats per min (Fig.2.10(a,b)) [88].

2.3.3. Micro-Structured Capacitive Pressure Sensors

Typically, capacitive pressure sensors with unstructured elastomeric dielectric layers are sandwiched between two flexible electrodes [89]. The pressure sensitivity of a capacitive pressure sensor is determined when the elastomeric dielectric layer deforms (i.e., reduction in the dielectric layer thickness) to an applied external pressure causing change in capacitance to measure the applied pressure. However, to detect weak low pressure human physiological signals, it is critical to enhance the pressure sensitivity of capacitive pressure sensors [90]. To satisfy this critical
requirement, various elastomeric dielectric layer based capacitive pressure sensors have been studied [91,92]. Among various elastomeric dielectric layers, PDMS was considered a promising choice of material as it possesses excellent flexibility, and biocompatible with human skin [93,94]. However, the PDMS has had limitations in detecting weak, low pressure signals due to the Young’s modulus (stiffness) [95]. Therefore, by reducing the Young’s modulus (stiffness) property of the PDMS dielectric layer, the pressure sensing performance of a capacitive pressure sensor can be improved [95]. To reduce the Young’s modulus of PDMS elastomeric layer, inducing air gaps/air voids has been an effective strategy [96]. Therefore, micro-structures, and porous structures have been fabricated on the surface of PDMS elastomeric layers and that can be elastically deformed for low pressure ranges due to presence of air voids [97].

S. Kang et al developed and compared the pressure sensing performance of unstructured and micro-porous structured PDMS and concluded that the micro-porous structured pressure sensors were highly sensitive when compared to the unstructured pressure sensor. The fabricated micro-porous structured pressure sensor is shown in Fig. 2.11(a,b). As shown in Figure 2.11(c), a sensitivity of 0.63 kPa⁻¹ was obtained for the micro-porous structured pressure sensor with a 6
μm pore size in the low-pressure range (1–10 kPa). This value is considerably higher than the sensitivity of 0.08 kPa\(^{-1}\) obtained for the unstructured pressure sensor. This enhancement in sensitivity of the porous structured pressure sensor is attributed to the micro-porous structure which induces air voids in the PDMS elastomeric layer, thereby reducing the mechanical stiffness of the material and making it more flexible and elastic for low pressure ranges [98].

2.4. Energy Harvesting: A Low-Cost Power Solution for Wearable Electronics

The rapid advances in the development of miniaturized electronics have led to a gradual reduction in the power consumption of the electronics. Figure 2.12 shows the summary of power consumption values of generic and miniaturized electronics. Power consumption values are in the range of 100 nW–10 mW [99]. Most current portable miniaturized electronics uses commercial rigid rechargeable batteries that needs to be periodically replaced and recharged, which causes

![Figure 2.12. The average power consumed by generic and miniaturized electronics [99]](image)
environmental concerns, poor user experience, and can cause physical pain to the user in some cases [100-102]. Portable and wearable electronics require small light-weight power sources that can be integrated with cloth fabrics, and accessories such as watches, eyeglasses, shoes, and bracelets for wearable healthcare monitoring [103,104]. Therefore, it is important to develop power sources with soft, flexible, bendable and stretchable aspects to provide sufficient electricity to the portable and miniaturized electronics. Micro-and nano-energy harvesting techniques are widely researched and are capable of generating electricity from ambient energy sources such as wind, solar, hydro and mechanical (human body movements) and can power-up electronics that consume power in the range of 100 nW – 10 mW [105]. In the following sections, the author focuses on the types of energy harvesting mechanisms and different types of energy harvesting techniques that are used to realize flexible and portable power sources.

### 2.4.1. Working Mechanisms of Energy Harvesting

Energy harvesting can be realized though three different working mechanisms:

1. **Piezoelectric**
2. **Triboelectric**
3. **Pyroelectric**

Piezoelectric and triboelectric energy harvesting are two important mechanisms for harvesting mechanical energy from water waves [106], human motion [107], vibrations [108], and wind [109]. Piezoelectric mechanism utilizes strain-induced piezoelectric polarization in piezoelectric materials to convert mechanical energy to electrical energy [110] whereas, triboelectric mechanism uses triboelectrification and electrostatic induction effects to convert mechanical energy to electrical energy [111]. Pyroelectric based energy harvesting mechanism converts external heat energy to electric energy [112]. The change in temperature of a pyroelectric material
modifies the positions of the atoms within the crystal structure which results in the change of polarization of the material. This polarization change gives rise to a voltage and thus generates electricity [112].

2.4.2. Types of Energy Harvesting

The author will now introduce the reader to different types of energy harvesting techniques such as wind, solar, wave/tidal and mechanical energy harvesting that are currently being intensively researched in the field of portable and wearable electronics application.

2.4.2.1. Wind Energy Harvesting

The rapid advances in flexible materials have paved the way for developing wind energy harvesters with flexible, small, lightweight and cost-effective manufacturing characteristics [113]. Q. Zeng et al proposed highly efficient breeze detectors which resemble the structure of a tree for wing energy harvesting. In this work, flexible PET and polytetrafluoroethylene (PTFE)
was used for device fabrication (Fig. 2.13 (a-c)) and an instantaneous power of 1.5 mW was obtained. The obtained power was sufficient to power-up 200 LED simultaneously [114].

2.4.2.2. Solar Energy Harvesting

Electrical energy can be generated from solar power by means of the photovoltaic effect. When certain semiconductor materials are exposed to sunlight, the electrons and holes of the semiconductor material are separated. This creates a potential difference across the load that generates electricity [115]. J. Chen et al demonstrated a wearable textile based solar energy harvester using a ZnO nanowire and PTFE polymer wire-based material. The energy harvester was fabricated with a size of 4 cm × 5 cm. In this work, the fabricated solar energy harvester was used to charge a 2 mF commercial capacitor up to 2 V in 1 minute, under ambient sunlight [116]. The fabricated solar energy harvester is shown in Figure. 2.14.

Figure 2.14. Solar energy harvesting (a) A photograph of the fabricated textile based solar energy harvester and (b) photograph of an electronic watch continuously powered by the fabricated textile based solar energy harvester [116]

2.4.2.3. Tidal Energy Harvesting

Oceans cover over 70% of the earth’s surface [117], which contain various kinds of clean energy such as the water wave energy, tidal energy and ultrasonic energy [118,119]. P. Chen et al proposed efficient harvesting of water wave energy by using liquid silicone spherical structures and conducting electrodes. Electricity was generated when the soft rolling spherical liquid
silicone moves back and forth between two electrodes driven by water waves. This work uses triboelectric working principle to generate electricity and obtained maximum power of 10 mW that can sufficiently power up 200 LEDs [120]. Figure 2.15 shows the schematic and the fabricated tidal energy harvester.

2.4.2.4. Mechanical Energy Harvesting

Mechanical energy harvesting utilizes ambient energy available in the environment and harvests ambient mechanical/kinetic energy to convert to electricity and are widely used to power wearable electronics [121]. As for portable, and wearable electronic devices, the human body itself is a tremendous energy storehouse involving heat, mechanical energy, and bio-chemical energy. Human body movements such as ankle motion, knee motion, elbow motion and shoulder motion are capable of producing an approximate power of 33.4 W, 36.4 W, 2.1 W, and 2.2 W respectively, whereas, blood circulation and respiration can produce up to 0.9 W and 1 W respectively [122]. Therefore, by harvesting energy from human motion, flexible power sources that are small, lightweight and flexible can be realized. Z. Li et al proposed a high-performance biomechanical energy harvester for harvesting human body movements like gentle hand tapping, hand elbow movements, and normal human walking. With an effective device area of 16 cm²
under gentle hand tapping and hand movements, the reported energy harvester can deliver a current and voltage output up to 110 µA and 540 V, respectively. This work also demonstrated powering 560 LEDs simultaneously with the generated output voltage (Fig. 2.16 (a-e)) [123].

2.5. Summary

This chapter provided a brief background and introduction to FHE, a relatively new method of fabricating FHE devices, different types of printing as well as the need for implementing laser etching process for the fabrication of FHE devices. This was followed by a review on wearable pressure sensors, and ambient energy harvesters.
CHAPTER III

MICRO-PYRAMID STRUCTURED CAPACITIVE PRESSURE SENSORS USING LASER-ASSISTED PATTERNING PROCESS

3.1. Introduction

Pressure sensing has been a major focus for applications in various fields including soft robotics [124, 125], smart phones and displays [126, 127], health condition monitoring [128,129], electronic skin [130-133], tactile and touch sensing applications [134]. Pressure sensors based on different transduction mechanisms including resistive, capacitive, piezo-resistive and piezoelectric have been reported [135-137]. However, capacitive pressure sensors are highly desired, when compared to other sensing mechanisms, due to its enhanced sensitivity, stability and repeatable response for pressures ranging from the low to very high regimes [138-144]. The commercial pressure sensors in use, are not conformal enough for being directly attached on the skin for monitoring physiological signals and body movements in real time due to its rigid structure. With advances in flexible hybrid materials and fabrication technologies, exploration to develop highly sensitive, flexible and conformable pressure sensors is highly desired. This is specifically important for direct application on the skin as a wearable sensor in health monitoring applications and for advancing non-invasive monitoring of physiological signals [145-147].

Although, capacitive pressure sensors have been developed for various applied pressure regimes, they have not been explored enough for low pressure sensing applications, specifically in the 0 to 10 kPa pressure range which corresponds to E-skin applications [148, 149]. Typically, the compact and portable capacitance-based pressure sensors do not offer the sensitivity required for the low pressure ranges due to limitations in terms of hardness, deformation/compression and thickness of the dielectric layer. Research is focusing on improving the sensitivity of pressure
sensors for low pressure ranges by developing polydimethylsiloxane (PDMS) based microstructured dielectric layers to increase its deformation and contact area [150-152]. Various microstructures including micro-cones [153], micro-sphere [154], micro-pyramids [155], micro-pillars [156], porous structures [157], nanoparticles [158], microspheres [159] and micro-array structures [160] have been widely studied. Among these, micro-pyramids have shown excellent sensitivity owing to its structural deformation compared to other structures [161,162].

Flexible pyramid micro-structure (PM) based dielectric layers are typically fabricated using master molds which have been patterned using conventional processes such as lithography [163-165]. Photolithography techniques include processes such as wet, dry etching, masking, and annealing techniques to pattern the master molds and has produced highly uniform microstructures with good resolutions and controlled aspect ratios [166]. However, these fabrication processes involve complex steps and are relatively time-consuming, expensive and require clean-room facilities. Microstructures are also being fabricated by bionic micropatterning methods which uses lotus leaves [167], banana leaves [168], sandpaper [169], insect wings and flower petals [170] as natural master molds due to their structured surface morphologies. However, the microstructures fabricated using natural molds have non-uniform surface microstructures and the aspect ratio cannot be controlled. To overcome these limitations, a laser-assisted engraving process, which is a subtractive manufacturing process that enables rapid prototyping of flexible sensors, was envisioned as a solution to fabricate master molds to obtain the PM-PDMS [171].

The laser assisted patterning process, which is a relatively new type of fabrication technique, uses a fiber and carbon dioxide (CO2) laser sources for etching materials. In this process, the parameters of the laser beam such as power, speed, frequency, focal length, spot size and laser
pulse per inch (PP-I) can be fine-tuned to control the resolution and the quantity of material etched in the x-, y- and z-directions. This process is relatively simple, easy to operate and fast with less preparation time [171-175]. The aspect ratio and the depth of the cavities in the master mold can be easily controlled by varying the power level, speed and PPI of the laser beam during the laser assisted patterning process.

In this work, PM-PDMS has been fabricated from master molds that were developed using laser assisted patterning process. Then, the novel highly sensitive capacitive pressure sensor was formed by sandwiching the PM-PDMS dielectric layer between two screen-printed silver (Ag) electrodes. The capability of the pressure sensor to detect varying applied pressures ranging from 0 to 10 kPa was investigated. In addition, the hysteresis effect was also examined for the fabricated pressure sensor and is presented later in this document.

3.2. Experimental

3.2.1. Chemicals and Materials

A 1.5 mm thick acrylic sheet from Polymershapes, USA was used for fabricating the master mold. A Sylgard® 184 silicone elastomer kit (A & B) from Dow Corning consisting of elastomer base and hardener was used to fabricate the dielectric layer. MG 7-9900 soft skin adhesive PDMS kit (A & B) from Dow Corning was used for attaching the electrode to the dielectric layer. Ag based AG-800 from Applied Ink Solutions, USA was used to fabricate the electrodes. A flexible 125 µm thick Melinex® ST506 PET was used as the substrate. Anhydrous 99.5 % pure isopropyl alcohol (IPA) (278475) from Sigma-Aldrich chemical company, USA was used to clean the acrylic master mold as well as the PET substrate.
3.2.2. Sensor Fabrication Process

A grid consisting of 127×127 μm square patterns was designed using AutoCAD™ and then imported to the laser machine (PLS6MW, 75 Watts, Universal Laser Systems, USA). The fabrication steps for a 35 mm diameter PM-PDMS pressure sensor is shown in Fig. 3.1(a-g). Initially, the surface of the acrylic sheet is patterned with the designed square patterns using a CO₂ laser beam with 10.6 μm wavelength, 15 % power and 5 % speed (Fig. 3.2(a)). The photograph of the fabricated master mold (M1) with cavities is shown in Fig. 3.2(b). The cavities formed on the acrylic sheet (Fig. 3.2(c)) due to the laser ablation was then cleaned with the IPA solution. The PDMS solution was prepared by mixing the elastomer base and hardener in a 10:1 ratio, for 3 minutes at 1000 rpm using an AR-100 Planetary Centrifugal Mixer from Thinky USA, Inc. Both elastomer base and hardener were mixed homogenously, and no bubbles were observed. The mixed PDMS was then poured on the engraved acrylic mold M1 (Fig. 3.2(d)) and thermally cured for 30 minutes in a VWR 1320 oven at 60 °C. After curing, the PM-PDMS dielectric layer

![Diagram of fabrication steps](image)

**Figure 3.1. Fabrication steps of the PM-PDMS based pressure sensor**
DL1 was peeled-off from the master molds M1 (Fig. 3.2(e)). Similarly, PM-PDMS dielectric layers DL2 and DL3 were fabricated using master molds M2 and M3 that were developed using the speed and power provided in Table 3.1. The height, h of the microstructures formed on the 1.5 mm thick, d of the PDMS dielectric layers DL1, DL2, and DL3 were measured using a high-resolution digital microscope (DinoLite Edge AM7915) (Fig. 3.3(a-c)). Another set of DL1, DL2 and DL3 were fabricated for assembling the pressure sensors. A total of 9 pressure sensors have been fabricated among which, first three sensors were fabricated with micro-pyramid height of 1500 μm, and next three sensors with a micro-pyramid height of 1000 μm, and other three sensors with a pyramid height of 750 μm.

Figure 3.2. Master mold fabrication: (a) laser patterning, (b) patterned acrylic master mold, (c) cross-section view of acrylic mold with patterned cavities; Pressure sensor fabrication: (d) PDMS poured on patterned acrylic (e) peel off PDMS with micro-pyramids

Figure 3.3. Fabricated micro-pyramid structured PM-PDMS dielectric layers (DL) with different micro-structure heights (a) DL1 with height of 1500 μm, (b) DL2 with height of 1000 μm, and (c) DL3 with height of 750 μm
Table 3.1. Laser engraving parameters of acrylic master mold

<table>
<thead>
<tr>
<th>Master mold</th>
<th>Laser parameters</th>
<th>Dielectric layer</th>
<th>Microstructure height (µm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Power (%)</td>
<td>Speed (%)</td>
<td></td>
</tr>
<tr>
<td>M1</td>
<td>15</td>
<td>5</td>
<td>DL1</td>
</tr>
<tr>
<td>M2</td>
<td>15</td>
<td>8</td>
<td>DL2</td>
</tr>
<tr>
<td>M3</td>
<td>15</td>
<td>11</td>
<td>DL3</td>
</tr>
</tbody>
</table>

The design for the electrodes was created using AutoCAD™ and the screen was fabricated at MicroScreen®. The screen was of stainless-steel material with mesh count 325, wire diameter 28 µm and an emulsion thickness of 13 µm. Then, the electrodes for the PM-PDMS pressure sensor (PS1) was fabricated by screen printing a 4.5 µm thick Ag ink on the PET substrate using an HMI 485 Semi-Automatic Screen printer. The printed sample was cured at 130 °C for 8 minutes. Following this, the top and bottom Ag electrodes were attached to the smooth side of the two dielectric layers DL1 (h=1500 µm) using adhesive PDMS. Finally, the pressure sensor (PS1) was obtained by placing the top and bottom electrodes with the PM-PDMS layers in such a way that the pyramid microstructures (PM) of both the top and bottom electrodes face each other, without the use of any adhesives. Similarly, pressure sensors PS2 and PS3 were fabricated by sandwiching

(a) 
(b) 

Figure 3.4. PM-PDMS based capacitive pressure sensor (a) top-view and (b) side-view of fabricated pressure sensor (PS1)
the dielectric layers DL2 (h=1000 μm) and DL3 (h=750 μm) between the top and bottom electrodes, respectively. The top and side view of the fabricated pressure sensor (PS1) is shown in Fig. 3.4(a) and (b).

3.2.3. Experiment Set-up

The experiment setup is shown in Fig. 3.5. The electrodes were connected to the jumper wires (SparkFun Electronics, PRT- PRT-11026) using contact crimp socket interconnects (TE Connectivity AMP Connectors, 2-487406-2). Then, the sensors were placed on a Mark-10 test stand platform and pressures ranging from 0 to 10 kPa were applied using a vertically movable...
force gauge (Mark 10 ESM 303). The capacitive response of the sensor towards varying pressures was measured using an Agilent E4980A precision LCR meter. A computer with custom-built LabVIEW program was connected to the LCR meter via USB and was used for post-processing and data analysis. All experiments were performed at room temperature.

3.3. Results and Discussion

3.3.1. Capacitive Pressure Sensor Response

The performance of all the 9 fabricated pressure sensors were investigated by measuring its

Figure 3.6. Capacitive pressure sensor response of the fabricated PM-PDMS pressure sensor (a) dynamic capacitance response of pressure sensor PS1, (b) PS2, (c) PS3, and (d) Relative Capacitance change of pressure sensors PS1, PS2, and PS3
The working of the pressure sensor is similar to that of a parallel plate capacitor and the capacitance is calculated using Eq. (1).

\[ C = \frac{\varepsilon_0 \varepsilon_r A}{d} \]  

where \( C \), \( \varepsilon_r \), \( \varepsilon_0 \), \( A \), and \( d \) are the capacitance, effective dielectric constant, permittivity of free space, device contact area, and the thickness of dielectric material, respectively. When an external pressure is applied on the sensor, the PM of PDMS dielectric material deforms or gets compressed.
Table 3.2. Sensitivities of pressure sensor PS1, PS2, and PS3

<table>
<thead>
<tr>
<th>Sensors</th>
<th>Sensitivities (% Pa⁻¹)</th>
<th>0 Pa to 100 Pa</th>
<th>100 Pa to 1000 Pa</th>
<th>1 kPa to 10 kPa</th>
</tr>
</thead>
<tbody>
<tr>
<td>PS1</td>
<td>Y1= 0.221</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PS2</td>
<td>Y4= 0.160</td>
<td>Y2= 0.033</td>
<td></td>
<td>Y3= 0.011</td>
</tr>
<tr>
<td>PS3</td>
<td>Y7= 0.099</td>
<td>Y8= 0.015</td>
<td></td>
<td>Y9= 0.003</td>
</tr>
</tbody>
</table>

(due to decrease in d) and the capacitance increases, which will be proportional to the applied pressure.

The capacitive responses of PS1, PS2 and PS3 were investigated for three different pressure ranges: 0 Pa to 100 Pa in steps of 10 Pa, 100 Pa to 1000 Pa in steps of 100 Pa, and 1 kPa to 10 kPa in steps of 1 kPa. The dynamic response of PS1, PS2, and PS3 are shown in Fig. 3.6 (a-c).

Initially, a base value of 3 pF was measured for the PS1 at 0 kPa (no applied pressure) for 10 seconds. Following this, a pressure of 0.01 kPa was applied using the Mark-10 force gauge and a corresponding capacitance of 3.1 pF was measured for 10 seconds. Then, the applied pressure was removed, and the capacitance value of the sensor (3 pF) was recorded for another 10 seconds.

These steps were repeated by increasing the applied pressures up to 10 kPa, and the corresponding capacitance values were recorded. An overall increase in capacitance from 3 pF (base value) to 7.4 pF was observed for the applied pressures, 0 to 10 kPa. This corresponds to a relative capacitance change of 153%. Similarly, relative capacitance changes of 89.9% and 65.7% were obtained for the PS2 and PS3 respectively (Fig. 3.6(d)). The sensitivities obtained for all the pressure sensors in the pressure range of 0 Pa to 100 Pa, 100 Pa to 1000 Pa, and 1 kPa to 10 kPa are shown in Figure 3.7 (a-c) with a standard deviation. Pressure sensor, PS1 with pyramid height of 1500 µm demonstrated the maximum sensitivity in all the three pressure ranges since the tip of the pyramid microstructure with larger height undergoes more deformation for applied pressure.
than the pressure sensors PS2 and PS3 (Table 3.2). Therefore, PS1 was chosen for further characterization and for demonstrating the application in this work.

3.3.2. Hysteresis Response

The hysteresis response of the PS1 was investigated for the pressure range of 1 to 10 kPa. Step-wise pressures increasing from 1 to 10 kPa and then decreasing from 10 kPa to 1 Pa, in steps of 1 kPa were applied with 10 cycles per step (Fig. 3.8(a)). The capacitive response of the pressure sensor was measured and the maximum hysteresis (MH) was mathematically calculated using Eq. (2) where $x_1$ and $x_2$ are the capacitances measured for an applied pressure during stepwise increase and decrease, respectively. $x_p$ is the peak capacitance at 10 kPa and $x_b$ is the base capacitance (at 0 kPa). An MH of 0.7% was calculated at an applied pressure of 9 kPa where $x_1$ and $x_2$ were measured to be 7.27 pF and 7.29 pF, respectively (Fig. 3.8(b)). This indicated that the PM-PDMS based pressure sensor has relatively better recovery and elasticity characteristics.

$$MH\% = \left[ \frac{x_2 - x_1}{x_p - x_b} \right] \times 100$$

(2)

![Figure 3.8. Hysteresis response of the fabricated PM-PDMS pressure sensor (PS1): (a) step-wise pressure response, and (b) hysteresis response](image)
3.3.3. Repeatability

The repeatability of the PS1 was investigated by subjecting it to 1000 cycles of pressure loading (loading - 20 seconds; unloading - 20 seconds; applied pressure - 5 kPa). Figure 3.9 shows the repeatability of the sensor response; the inset shows the capacitive response of the sensor for 10 cycles between 200-210 and 800-810 at applied pressure of 5 kPa. It was observed that the capacitance response of the pressure sensor over the 1000 cycles increased from a base value of ≈3.0 pF to ≈6.4 pF, demonstrating a maximum change of 112.0 ± 1.3%. Thus, it can be concluded that the performance of the sensor is highly repeatable and durable.

3.3.4. Response and Recovery Time

The response and the recovery time of the PS1 was investigated by subjecting it to an applied pressure of 20 Pa (Fig. 3.10(a)). It was observed that the capacitance response of the pressure sensor increased from a base value of 3.03 pF to 3.13 pF, following the response pattern of the force gauge. The response time of the pressure sensor, defined as the time taken for the

![Figure 3.9. Repeatability of PM-PDMS pressure sensor (PS1)](image)

Figure 3.9. Repeatability of PM-PDMS pressure sensor (PS1)
capacitance to reach 90% \((T_{90\%} = 2.85 \text{s})\) of the total capacitance change \((0.1 \text{ pF})\) from 10% \((T_{10\%} = 2.80 \text{s})\), was calculated to be 50 ms. Then the time taken by the force gauge to reach from 10% \((T_{10\%} = 2.70 \text{s})\) to 90% \((T_{90\%} = 2.80 \text{s})\) of the total applied pressure of 20 Pa was calculated to be 0.1 s. Similarly, a recovery time of 150 ms was calculated for the pressure sensor, when the applied pressure of 20 Pa was removed (Fig. 3.10(b)). A comparison summary, of the present work with previously reported pressure sensors, is shown in Table 3.3. It can be noticed that the PM-PDMS pressure sensor has an appreciable response time and sensitivity when compared to the reported pressure sensors.
To demonstrate the practical applicability of the fabricated pressure sensor (PS1), activities such as hand gestures and carotid pulse measurement of a healthy female subject was monitored and recorded. Initially, the PS1 was mounted on the wrist of a subject and its capacitance response for monitoring hand closing and opening gesture (Fig. 3.10(a)). A base capacitance of 4.85 pF was measured when the fingers of the subject’s hand was closed. It was observed that the capacitance increased from 4.85 pF to 5.5 pF when the subject opened the fingers of the hand. Similarly, when the fingers were closed, the capacitance decreased to 5.0 pF. This results in a relative capacitance change of 14% during the hand closing and opening gestures, which
corresponds to a pressure of 50 Pa. The capacitance change can be attributed to the deformation of the micro-pyramids during the opening and closing (of the fingers) of the hand.

Following this, another PM-PDMS based pressure sensor of dimensions 2.5 × 1.5 cm was fabricated for monitoring the carotid pulse of a healthy female subject. The sensor was attached to the neck of the subject using adhesive PDMS and flexible conductive copper tape (Bertech CFT-2, 2 mil thick). The results show 20 cycles of wave pulses of the female subject (Fig. 3.10(b)). The waveforms were normal and repeatable for 85 beats per minute. The results obtained demonstrated the capability of using laser-assisted patterning for fabricating PM-PDMS based flexible pressure sensors for implementation in low pressure sensing applications.

3.4. Summary

In this chapter, the author presented an introduction that discusses the importance of developing a flexible micro-structured pressure sensor for wearable health monitoring. A detailed account of the experimental tasks involved in this work was then presented. This includes the
chemicals, materials and devices used and experiment setup used. Finally, the results obtained have also been presented.

To summarize, capacitive pressure sensors were fabricated using laser-assisted patterning. The micro-pyramid structured pressure sensors were fabricated with PM-PDMS and Ag as the dielectric and electrode layers, respectively, on flexible PET substrate. PM-PDMS films were fabricated using a laser engraved acrylic mold. Ag was screen printed on the PET film to form the top and bottom electrodes. The capability of the PM-PDMS pressure sensor was investigated by subjecting it to pressures ranging from 0 to 10 kPa. Sensitivities of 0.22% Pa$^{-1}$, 0.03% Pa$^{-1}$ and 0.01% Pa$^{-1}$ along with a correlation coefficient of 0.9536, 0.9586 and 0.9826 were obtained for the pressure sensor in the pressure range of 0 Pa to 100 Pa, 100 Pa to 1000 Pa, and 1 kPa to 10 kPa, respectively. The use of the flexible PM-PDMS enabled the pressure sensor to exhibit salient features including a fast response time of 50 ms, low hysteresis of 0.7%, and a rapid recovery time of 150 ms as well as excellent repeatability and durability over a wide pressure range. The results demonstrated the feasibility of employing laser-based patterning techniques for the development of cost efficient, flexible and highly repeatable pressure sensors. These sensors can be employed for several E-skin and soft robotic applications because of their stable performance in the pressure range of 0 to 10 kPa.

In the following chapter, the author presents a project that involved the development of a triboelectric energy harvester for providing a flexible power source for FHE devices. The author discusses the design, fabrication, and characterization of the triboelectric energy harvester. The experiment set-up, testing and results obtained are also presented.
CHAPTER IV
TRIBOELECTRIC NANOGENERATOR (TENG)

4.1. Introduction

Energy harvesting has been a major focus of research in recent years for providing low-cost power solutions for flexible hybrid electronics (FHE) [176,177]. FHE typically consists of soft, flexible and stretchable devices for next-generation electronics, and the challenge is to provide flexible power sources to drive these FHE based devices [178,179]. Energy harvesting techniques takes advantage of ambient energy sources such as wind, solar, hydro, tides and biomass, and utilizes different transduction mechanisms to generate electricity. Nanogenerators are sustainable power sources, which can effectively harvest mechanical energy [180,181]. Nanogenerators based on different transduction mechanisms including electromagnetic [182,183], electrostatic [184-186], piezoelectric [187,188] and triboelectric [189-191] have been reported. Among these, the triboelectric based nanogenerators (TENG) are highly desired due to their high efficiency, low cost and high reliability for both high power and low power electronics [192]. TENGs operate in four fundamental modes: vertical contact-separation, lateral sliding, single-electrode and free-standing. The vertical contact-separation [193,194] and single-electrode [195,196] modes are based on the mechanism of contact electrification. The lateral sliding [197,198] and free-standing [199,200] modes are based on sliding electrification mechanism. In all these four modes, charge transfer is induced and a potential difference across the electrodes is generated, when they are mechanically excited. However, the vertical contact-separation and lateral sliding modes typically use two electrodes (working and counter electrodes), whereas, the single-electrode and free-standing modes use one working electrode with a reference electrode. For energy harvesting applications, the vertical contact-separation mode has demonstrated relatively higher output
voltages (200 V) [201]. Therefore, the development of novel TENGs that operate in the vertical contact-separation mode is important.

TENGs generate static electricity between two materials with excess positive and negative charges on their surfaces. The static electricity depends on the intrinsic properties, surface charge densities and surface morphology of the TENG polymer based triboelectric layer [202-204]. Different microstructure-based surface morphologies for the triboelectric layers have been fabricated using master molds created with photolithographic patterning [205-208] and printed circuit boards [209-212]. However, these fabrication processes are relatively time-consuming, involves complex steps and require expensive equipment. Laser-assisted patterning [213] is a facile-method of fabricating molds, for creating microstructures, with resolution that is like that of traditional photolithographic based microstructures.

In this work, a vertical contact-separation mode based triboelectric nanogenerator was developed. Polydimethyl siloxane (PDMS) and Kapton were chosen as the positive and negative triboelectric layers, respectively. The PDMS based microstructures were fabricated using master molds created with laser-assisted patterning. The capability of the TENGs to generate electricity was investigated by applying mechanical vibrations, for varying frequencies. The details of the fabrication process and the electromechanical response of the TENG is analyzed and presented in this paper.

### 4.2. Working Theory of TENG

The working principle of the TENG device is shown in Figure. 4.1. Initially, when the device is at rest, there is no charge transfer between the two electrodes (Fig. 4.1(i)). When the device is mechanically excited by pressing it together (Fig. 4.1(ii)), the MP-PDMS loses electrons and
becomes positively charged. Whereas, Kapton gains electron and becomes negatively charged. Now, the device is fully pressed and there is no further flow of electrons, as the electrostatic equilibrium has been achieved (Fig. 4.1(iii)). When the distance between Kapton and MP-PDMS increases due to separation, when the mechanical force is released, the electrons are transferred from the Kapton to the MP-PDMS (Fig. 4.1(iv)).

Figure 4.1. Working mechanism of triboelectric nanogenerator (TENG)
4.3. Fabrication of Triboelectric Nanogenerator (TENG)

4.3.1. Chemicals and Materials

A flexible polyimide film (DuPont™ Kapton® 500HN) was used as the substrate. A soft skin adhesive kit (MG 7-9900) and Sylgard® 184 silicone elastomer kit (with elastomer and hardener) from Dow Corning was used to fabricate the triboelectric layer. An acrylic sheet (1.5 mm Polymershapes, USA) was used for the master mold. A silver (Ag) ink (AG-800) from Applied Ink Solutions, USA was for the electrodes. Anhydrous pure isopropyl alcohol (IPA) (99.5 %, Sigma-Aldrich chemical company, USA) was used as the cleaning solvent.

4.3.2. TENG Fabrication Process

The fabrication steps for the micro-pyramid structured TENG (MP-TENG) is shown in Fig. 4.2. First, the acrylic sheet was cleaned with IPA (Fig. 4.2(i)). Then, a grid design consisting

![Fabrication process of TENG](image)

**Figure 4.2. Fabrication process of TENG**
of square patterns (200×200 μm), designed using AutoCAD™, was engraved using a laser machine (PLS6MW, 75 Watts, Universal Laser Systems, and USA). A CO2 laser beam with 10.6 μm wavelength, 15% power and 5% speed were used for the laser patterning (Fig. 4.2(ii)).

Following this, the PDMS based triboelectric layer was prepared by mechanically mixing the elastomer base and hardener in a 10:1 ratio, for 10 minutes. The mixed PDMS was then poured on the engraved acrylic mold (Fig. 4.2(iii)), and thermally cured for 30 minutes in a VWR 1320 oven at 60 °C. After curing, the PDMS layer, with the micro-pyramid structures (MP-PDMS), was peeled-off from the mold (Fig. 4.2(iv) and Fig. 4.2(v)). The electrodes for the TENG were fabricated by screen printing (HMI 485 Semi-Automatic Screen printer) the Ag ink on the Kapton substrate. The peeled off MP-PDMS layer was attached to the bottom electrode using the adhesive soft skin PDMS and thermally cured in an oven at 45°C for 10 minutes. Finally, the bottom electrode with MP-PDMS was assembled with the top electrode, where the two triboelectric layers of MP-PDMS and Kapton face each other (Fig. 4.2(vi)). The top and side view of the fabricated MP-TENG is shown in Fig. 4.3(a and b).
4.3.3. Experiment Set-up

The experiment setup is shown in Figure 4.4. A function generator was connected to a shaker (Vibration Test System, Model No. VG 100-8), via a power amplifier (Techron, Model No. 7705). The fabricated MP-TENG was mounted with its bottom electrode attached to the shaker using a custom-built adjustable clamp. The top electrode was mounted on a thick glass plate using a G-clamp and placed on the MP-PDMS. The electrodes of the TENG were connected to a digital oscilloscope (Tektronix TDS5104B Digital Phosphor Oscilloscope) for monitoring the voltage response of the fabricated TENG. The shaker was used to apply sinusoidal vibrations on the TENG with frequencies ranging from 5 Hz to 40 Hz and a constant amplitude.

![Experiment set-up diagram](image)

**Figure 4.4. Experiment set-up**
4.4. Results and Discussion

To determine the operating condition of the fabricated MP-TENG, the open circuit voltage and short circuit current for the power generator device was investigated by connecting a resistive load across the MP-TENG. Figure 4.5 shows the voltage and current response of the MP-TENG for resistive load varying from 0.35 kΩ to 15000 kΩ. It was observed that the voltage increased from 0 V to 4.3 V and the current decreased from 6 μA to 0 μA as the resistive load decreased from 0.35 kΩ to 15000 kΩ, respectively. A maximum open circuit voltage of 4.3 V was obtained at the maximum load resistance of 15000 kΩ. In addition, the maximum short-circuit current of 6 μA was observed at a negligibly small resistance of 0.35 Ω, indicating that no load is connected to the device. Using the open circuit voltages and short circuit currents, the instantaneous power of the TENG for the varying resistive loads (0.35 kΩ to 15000 kΩ) were calculated. Figure 4.6 shows the calculated instantaneous power of the TENG and it was observed that the maximum output power was 2.5 μW at 1000 kΩ load resistance.

![Graph showing voltage and current response of TENG](image)

**Figure 4.5. Open-circuit voltage and short-circuit current characteristic of MP-TENG**
The power generation capabilities of the fabricated MP-TENG was demonstrated by investigating the open circuit voltage, which represents the maximum voltage that the MP-TENG can provide to an external circuit. Figure 4.7 shows the open circuit voltage stabilizing at 4 Vp-p, for frequencies varying from 5 Hz to 40 Hz. The open circuit voltage remained unchanged and stable for the increasing frequencies because the total number of charges transferred between the triboelectric layers at any given time was constant.
4.5. Summary

In this chapter, the author introduced a novel triboelectric nanogenerator based energy harvesting device. A brief background and introduction about the need for triboelectric nanogenerator was provided. The triboelectric nanogenerator was then fabricated using laser-assisted fabrication process. A detailed account of the experimental tasks involved in this work was presented. This includes the chemicals and materials used, fabrication process and the experiment set-up used. Finally, the capability of the fabricated triboelectric nanogenerator was investigated by measuring the open-circuit voltage, short-circuit current, and instantaneous power delivered to the varying load resistance.

In summary, A flexible TENG was fabricated using laser-assisted patterning and additive screen-printing processes. The MP-PDMS was prepared by using a laser engraved acrylic mold. A top and bottom electrode for TENG was fabricated by screen printing Ag ink on a flexible polyimide substrate. MP-PDMS and the printed electrodes were chosen as the positive and negative triboelectric layers, respectively. The performance of the fabricated MP-TENG was improved by modifying the surface morphology of the polymer material. It was observed that the open-circuit voltage was stabilized at 4 V with a short-circuit current of 6μA resulting in an instantaneous power of 2.5μW and a power density of 0.181 μW/cm².

In the following chapter, the author concludes this dissertation with a summary of the project performed and provides some suggestions for possible future work.
CHAPTER V
CONCLUSION AND FUTURE WORK

5.1. Conclusion

The author has successfully demonstrated the design, fabrication and characterization of micro-pyramid structured, and M-Tooth hybrid structured based capacitive pressure sensor for wearable health monitoring and a triboelectric nanogenerator by using laser-assisted patterning technique. The thesis was organized and pursued in two projects in order to achieve the research outcomes.

In the first project, capacitive pressure sensors with micro-pyramid structure was successfully fabricated and tested. The micro-pyramid structured capacitive pressure sensor was fabricated using laser-assisted patterning of master mold and electrodes were fabricated by depositing Ag on flexible PET using screen-printing process. Characterization of micro-pyramid structure heights were performed. Sensitivities of 0.22% Pa\(^{-1}\), 0.03% Pa\(^{-1}\), and 0.01% Pa\(^{-1}\) along with a correlation coefficient of 0.9536, 0.9586 and 0.9826 were obtained for the pressure sensor in the pressure range of 0 Pa to 100 Pa, 100 Pa to 1000 Pa, and 1 kPa to 10 kPa, respectively. A maximum hysteresis (MH) of 0.7% was calculated at an applied pressure of 9 kPa indicating that the pressure sensor has a good recovery and elasticity characteristics. The fabricated pressure sensor was also highly repeatable and durable for 1000 cycles of loading and unloading. The hand gestures and carotid pulse measurement of a healthy female subject was monitored and recorded by the fabricated pressure sensor which resulted in the relative capacitance of 14% for hand gesture monitoring and 2.2% for carotid pulse measurement. The author thus demonstrated the feasibility of employing the fabricated pressure sensors in E-skin and soft robotic applications. The outcome of this project was published in [214, 215].
In the second project, a triboelectric nanogenerator (TENG) was fabricated and tested. Microstructured polydimethyl siloxane (PDMS) and Kapton were chosen as the positive and negative triboelectric layers, respectively. The microstructured PDMS was fabricated using master molds created with laser-assisted patterning. Electromechanical response of the fabricated TENG with respect to low-frequency vibrations were analyzed. The TENG generated a maximum peak to peak open-circuit voltage of 4 V for frequencies varying from 5 Hz to 40 Hz with a short-circuit current of 6 μA. An instantaneous power of 2.5 μW was obtained with a power density obtained of 0.181 μW/cm², at 15 MΩ load resistance. Thus, the fabricated TENG with a power density 0.181 μW/cm² was sufficient to power small scale electronics. By further enhancing the power density of TENG, it is possible to replace rechargeable batteries and thus can serve as portable power source for future FHE devices. Further research is underway to enhance the power output of TENGs by fabricating TENGs that can be integrated on to a textile for human motion energy harvesting. The outcome of this research project was published in [216]

5.2. Future Work

The author believes that there are few areas to improve the current projects. Some suggestions for the future work are discussed below.

**Micro-Structured Capacitive Pressure Sensor for Wearable Health Monitoring:**

- To have tunable sensitivities of the micro-pyramid structured pressure sensor for higher dynamic pressure ranges.
- Fatigue testing and analysis of micro-pyramid structures are required to know the maximum mechanical bending, compression, and deformation of the micro-pyramid structures.
➢ Investigation of various aspect ratio of micro-pyramid structures to optimize the sensitivities of the pressure sensors for wide pressure ranges.

**Triboelectric Nanogenerator:**

➢ To harvest low-frequency vibrations using an energy harvester, triboelectric material’s surface morphology could be modified by creating different micro-patterns that can make the triboelectric layer more deformable for low-frequency vibrations.

➢ To study the surface charge densities of different triboelectric material for higher output voltage and power generation.
REFERENCES


