Chapter 7

Graphite-Polydimethylsiloxane Sensor

Abstract This chapter represents the fabrication and implementation of a novel sensor prototype developed from 3D printed mold-based technique. 3D printing was done to develop molds of specific dimensions where the casting of Graphite and PDMS was done to develop the electrodes and substrates of the sensor patches respectively. The developed sensors were used for monitoring strain-induced physiological movements by attaching them to different joints of the body. They were also employed for low-force sensing where identical objects with different weights were positioned in the sensing area of the patches to determine their responses.

7.1 Introduction

This chapter presents the development and implementation of novel Graphite-PDMS sensors for different strain-sensing applications. The fabrication of the sensor patches was done via 3D printed molds which were formed using acrylonitrile thermoplastic polymer as the printing filament. One of the enhanced techniques to fabricate sensors with flexible substrates has been the utilization of 3D printing technique where the master mold can be printed with specified dimensions quickly at a very low price, followed by their deployment to develop the final sensor patches. Another big advantage of using 3D printing for fabrication purposes is the usage of thermoplastic polymers to form the templates. Some of the advantages of these polymers lie in their high tensile strength, easy bendability, easy recyclability and high performance in terms of fatigue properties in comparison to metals (Advantages of thermoplastics), which makes these thermoplastic polymers (Kanoun et al. 2014) cheaper and much-preferred choice for 3D printing purposes. Graphite and PDMS were considered to fabricate the

electrodes and substrates of the sensor patches respectively. The merits of using PDMS over the other commonly used polymers are its low cost, high tensile strength, hydrophobicity, resistance towards temperature and ability to form excellent interfacial bonding with the polymer matrix (Nour et al. 2012). The advantages of using graphite lie in its high compressive strength, high electrical and thermal conductivity and corrosion resistance. One of the biggest merits of using graphite is its biocompatibility (Chang-bin et al. 2011), which makes it a preferred choice over other conductive materials to fabricate the devices for biomedical applications. The electrodes were formed in an interdigital manner via the casting of graphite powder on the trenches of 3D printed molds having specific dimensions. The working principle of the sensor patches is explained along with the COMSOL simulation result which depicts the electric field density distribution between the two electrode fingers of opposite polarity under applied stress. The characterization of the sensor patches was done to analyze their responses to the change of frequency and applied stress. The experimental results sensor patches were obtained by determining the changes in their complex conductivities and capacitances at different operating conditions. The sensor patches were then deployed for two applications. One was regarding strain-induced physiological parameter monitoring by attaching them to different parts of the body like finger, elbow, neck, and knee. Strain sensing was done based on the bending of the sensor patches along with the bending of the different joints. The other application was related to low-force sensing measurements. The results depicted by the developed sensor patches were promising, thus increasing the chances of utilizing them in future for dynamic fields like biomedical field, microfluidic and tactile applications. The merits of the proposed sensor patches over the existing methods include its low-cost of fabrication, low-cost of processing materials, simple operating principle and a wide range of strain-induced applications. The proposed sensor patches also have lightweight and small dimension, which makes them easier to be replaced or fixed if damaged. The sensors had a single-sided measurement with the electrodes being capacitive in nature, thus making them easier to be adjusted and fabricated while achieving high sensitivity with little stress. The input power required is also low but obtaining a good resolution and

response to tested frequency ranges. Apart from utilizing the flexible nature of these sensor patches, they can also be employed for monitoring non-metallic targets and dielectric materials.

7.2 Fabrication of the Sensor Patches

The fabrication of the sensor patches was done inside the laboratory conditions. Figure 7.1(a) shows the 3D printing system that was used to fabricate the molds. Acrylonitrile Butadiene Styrene (Tymrak et al. 2014) was considered as the printed filament to form the moulds. This material was considered because of high flexural strength and ductility in comparison to other thermoplastics used for 3D printing purposes.

Figure 0.1 The 3D printing systems (a) were used to develop the molds (b) using the Acrylonitrile Butadiene Styrene as the printing filament. The height of the trenches on the mold was adjusted to 500 microns.

The filament was attached to the printing device (3D PRINTING SYSTEM) followed by pre-heating the later for fifteen minutes before starting the process. This was done to let the device to achieve a fixed temperature required for the operation. This was followed by

uploading the designs of the electrodes that are to be fabricated on the sensor patch using the designing software CREO Parametric 2.0, that was associated with the printing system. The height of the trenches to be developed on the molds was adjusted to 500 microns. The mold formed by the printing system is depicted in Figure 7.1(b). After the molds were developed, they were thoroughly washed using 2-isopropanol to remove the presence of any kind of snippets of the printing filament. Graphite powder (Sigma-Aldrich 282863-25G, $<20\ \mu\text{m}$) was then cast carefully upon the trenches of the mold as shown in Figure 7.2(a), to form the electrodes of the proposed sensor patch. An optimization on the volume fraction of the graphite powder used was done prior to the development of the electrodes via changing its critical volume fraction to vary the electrical conductivity and flexibility of the sensor patches.

Figure 0.2 Graphite powder was cast on the 3D printed mold (a), followed by the casting of PDMS (b) on top of it. These two layers defined the electrodes and substrate of the sensor patches.

Since the homogenous weight fractions and volume fractions can be approximated to the same for a uniform and homogenous system (Malekie and Ziaie 2015), the critical volume fraction of the graphite powder to PDMS was optimized to 4.38% as shown in Figure 7.3.

The residual graphite powder was scraped off the mold to avoid short-circuital tendencies between the electrodes. The casting of the graphite powder was followed by the casting of a layer of PDMS (SYLGARD® 184 SILICONE ELASTOMER KIT) on the top of it as shown in Figure 7.2(b), which was formed by mixing the base elastomer and curing agent at a ratio of 10:1. The height of the PDMS on the mold was adjusted to around 1000 microns to form the substrate of the sensor patches on the 3D printed moulds with a casting knife (SHEEN, 1117/1000mm). This was followed by the desiccation of the sample for an hour to remove the trapped air bubbles trapped in it. Then the sample was cured in the oven at 70 °C for 2 hours. The curing of the PDMS solidified to form the substrate of the sensor patches. Figure 7.4 depicts the dimensions of the developed sensor patch.

Figure 0.3 Optimisation of the critical volume fraction of the graphite powder to develop the electrodes of the sensor patch.

The electrodes were designed in an interdigitated manner with three pairs of fingers for each electrode. The total surface area (A) of the sensor was fixed to 900 mm² with a length of 18 mm and width of

2 mm for each electrode finger. The interdigital distance (d) between two consecutive electrode fingers was fixed to 1.5 mm. The dimensions of the surface area of the sensor patches are critical for the chosen application as the effective change in the dimension of the sensor patches reflected the induced strain. The number of electrodes fingers on the sensing surface have been increased to cause a net increase in the generated electric field (Zia et al. 2011). The dimensions of the length and width of the electrodes were made big enough to cause an effect of the induced strain when attached to different parts of the body. The advantages of these fabricated sensor patches in terms of biomedical aspects lies in their biocompatibility due to the individual aspects of the processed raw materials (Chang-bin et al. 2011; Kim et al. 2011b).

Figure 0.4 Front view of the developed sensor patch. Three pairs of electrode fingers were present, each with a length and width of 18 mm and 2 mm respectively. The total surface area of the sensor patch was 900 mm².

Some of the merits of forming small sizes of the sensor patches are their easy reproducibility and replacement, low cost of production for sensors generated on a large scale, low input power requirement and improved sensitivity and accuracy in their responses. The interdigital nature of the electrodes of these sensor patches caused an easy operating principle during experimentation. The developed sensor patches after fabrication were again thoroughly cleaned with 2-isopropanol before using them for characterization and experimental purposes. Figure 7.5 depicts a comparison between the sizes of the developed patch with that of 50 cents of Australian coin to illustrate the small dimension of the formed sensor patch. Figures 7.6(a)-7.6(c) show the Scanning Electron Microscope (SEM) images of the top-view and cross-sectional views taken of the developed sensor patches taken with a Phenom XL. It is seen from Figure 7.6(a) that the PDMS layer came off smoothly and cleaned on the top of the graphite electrodes. The top view of the electrodes as shown in Figure 7.6(b), shows the mixture graphite and PDMS forming the electrodes. The cross-sectional view of the electrodes as shown in Figure 7.6(c) was formed when the sensors were peeled off from the 3D-printed mold after the solidification of PDMS.

Figure 0.5 Comparison between the sizes of the developed sensor with a 50 cent Australian coin.

Figure 0.6 SEM images of the developed graphite-PDMS sensor depicting the top-view of (a) PDMS, (b) graphite-PDMS mixture and (c) cross-sectional view of the electrodes.

7.3 Frequency Response and Stress-Strain Measurements

The analytical study of the responses of the sensor patches was regarded to their strain-induced nature. The illustration of the flexibility of the sensor patches is shown in Figure 7.7. Figures 7.8(a) and 7.8(b) depict the microscopic images taken with a DYNASCOPE microscope and a TOUPCAMTM (UCMOS03100KPA) camera to show a reference point with respect to no-strain and full-strain conditions. It is seen from the image that the strain applied on the sensor patch caused a horizontal movement with respect to its reference spot. The strain was exerted horizontally over the sensor patch by clamping one side of the patch. The changes in the capacitance values as a function of frequency were analyzed for four different chosen bending radii (1.5 mm, 3 mm, 4.5 mm, and 6 mm). These values were preferred to cover the range of the bending curvature for situations of the slightly bent situation to the highly bent. A frequency sweep was done between 1 Hz and 10 kHz at different bending radii of the sensor patches to determine the operating frequency as shown in Figure 7.9 for experimental purposes.

Figure 0.7 Image of the developed sensor patch depicting its flexibility.

Figure 0.8 Change in the reference spot in the sensor patch to depict their flexibility from (a) no- strain to (b) full-strain condition.

The stress-strain measurements were then performed to determine Young's modulus (E) of the sensor patches by applying tensile stress

to analyze the resultant strain. It is seen from the inset of Figure 7.10 that the tensile stress was applied to a direction which is perpendicular to the electrode fingers of the sensor patches. Figure 7.11 shows the stress-strain analysis of the developed sensor patches. The load was exerted on the sensor patch by clamping one side of the patches while vertically moving the other one upwards. The response of the sensor patches was within the elastic limit of the load-extension curve, and the extension followed linearity with the applied load for a high regression coefficient. The Young's Modulus (E) of the sensor patches at an extension of 6 mm was found out to be 675 kPa, calculated from the given values of stress (135 kPa) and strain (0.2). The linearity in the response of the sensor patches was validated with a high coefficient of determination (R^2). The change in capacitance values was also studied for the same values of the applied strain. The change in response to the patches due to the applied stress was also linear as shown in Figure 7.12.

Figure 0.9 Response of the sensor for the frequency sweep done between 1 Hz to 10 kHz at four different bending radii (1.5 mm, 3 mm, 4.5 mm, 6 mm) to determine the operating frequency.

Figure 0.10 Analysis of the stress-strain measurement. The direction of the applied stress shown is in the inset of the figure.

Figure 0.11 Response of the sensor patch in the proportional limit for the stress-strain measurement.

Figure 0.12 Response of the sensor patch for the change in capacitance occurring due to the applied strain.

The responses of the patches changed as a result of the extension generated in the horizontal direction of the sensor patches led to a change in the dimensions of the sensor patches in terms of their effective area (A') and effective interdigital distance (d'). The sensor patches extended from the normal position to the highest position of 10.2 mm before reaching the breaking point. The testing for the bending response of the sensor patches was done via bending them manually so that the electrodes were exerted under compressive load. Figure 7.13 depicts the change in capacitive values for the change in bending radii of curvature ranging between 6 mm and 11 mm. The capacitive values decreased with respect to the increasing radius. As the condition of the sensor patch was changed from normal to flexed state and vice versa, the dimensions of the sensor patch simultaneously oscillated between the normal and compressed state respectively. Figure 7.14 depicts the changes of the capacitive values for the normal and bending states of the sensor patch for continuous 10 cycles. The sensor was bent to a small radius of curvature and brought back to the reference position.

Figure 0.13 Capacitance of the sensor for a bending radius from 11 mm to 6 mm.

Figure 0.14 Response of the sensor for the experiments bending the sensor patch in a cyclic manner. The bent state of the sensor patch caused a decrease in its conductivity value. The inset of the figure shows the (a) normal and (b) bent states.

When the sensor patch changed from normal to bent state, the capacitance values simultaneously decreased from an average value of 35 pF to a value of 2 pF. This took place as a result of the increase in the radii of curvature due to the bending the patch which decreases the corresponding capacitance value. The response is shown in Figure 7.13. This attribute is advantageous in terms of the application of the sensor patches where they could be employed for monitoring the movement of a person (Nag et al. 2016b), low-force sensing (Nag et al. 2018) or for tactile measurements of prosthetic limbs (Nag et al. 2016a) based on the strain-induced nature of the electrodes. However, the range in the capacitance values for every bending cycle was not the same as a result of the re-orientation of the conductive fillers in the PDMS caused with every cycle. The relative resistance values of the sensor patches were calculated based on the ratio between the changes in corresponding resistance with a strain to the resistance values in its normal or reference condition.

Figure 0.15 Response of the sensor patch in terms of relative resistance with strain.

Figure 7.15 depicted the performance of the sensor patches in terms of the relative resistance with respect to the applied strain. The changes in values were analyzed by fitting it to a polynomial curve

with a power of 2. The response of the sensor patches adhered to the response of the graphite-based strain sensors (Tadakaluru et al. 2014) as seen from the experimental curve being very close to the fitted curve.

7.4 Strain Sensing

The deployment of flexible sensors has held an important role in the field of biomedical applications (Chen and Chatterjee 2013) for the last two decades. Sensors have been developed with a range of materials and subsequently characterized and implemented for various biomedical applications. There are some pressure sensors available commercially that could be used as strain and force sensors as shown in Table 7.1. These sensors can be used for various applications in the field of biomedical applications. The cost of fabrication, the dimension of the sensors and the complexity in the nature of the sensors are some of the attributes that affect their overall prices. Strain sensing has been among the popular choices of application where flexible electronic systems have been largely deployed (Hu et al. 2010; Stassi et al. 2014; Xiao et al. 2011) to analyze the tension caused as a result of the applied stress.

Table 0.1 Comparative study between the flexible sensors commercially available in the market.

Sl. No.	Sensor	Size (mm)	Price (\$)	Application
1	Flexiforce A201	50.8 x 9.7	69.00	Tactile sensors, human symbiotic robots
2	SEN-11207	56.8 x 25.4	47.00	Strain gauges
3	RB-Phi-121	25*11	45.00	Pressure sensors, touch user interface
4	Flexiforce A201-25	58.4 x 17.8	37.35	Touch pads, alarm systems
5	Flexiforce A101	15.6 x 7.6	34.00	Bed monitoring and force sensitive systems

Table 0.2 Comparative study between the strain sensors developed with different fabrication techniques.

Sl. No.	Fabrication technique	Size (mm)	Materials used	Application	Reference
1	Chemical vapor deposition	60 x 60	Graphene-PDMS	Facial expression, human motions	(Wang et al. 2014)
2	Thermoplastic processing technology	20 x 20	Carbon black-Thermoplastic elastomer	Body postures and body movements	(Mattmann et al. 2008)
3	Screen printing	2 x 2	Silver nanowires-PDMS	Thumb and knee movement	(Yao and Zhu 2014)
4	Encapsulation and molding	150 x 1.8	Copper-PDMS	Implantable medical device	(Kim et al. 2011b)
5	Encapsulation	110 x 80	RF transmitter-PDMS	Body movements	(Cheng and Wu 2011)
6	Dip-coating	40 x 40	Silver nanowires-PDMS	Pressure sensing	(Ge et al. 2016)
7	Sputtering	75 x 50	Copper-PDMS-Kapton	Body movements	(Yang et al. 2015)
8	Laser cutting	85 x 40	Carbon black-PDMS	Body sensing	(Weigel et al. 2015)
9	Chemical vapor deposition	50 x 10	Carbon Nanotubes-glass fiber	Health monitoring	(Sebastian et al. 2014)
10	3D printing	30 x 30	Graphite-PDMS	Body movements	Our work

Table 7.2 gives a comparative study for the performances of the proposed sensor with some of the research work done on similar sensors by other researchers with different methods. The comparisons are made in terms of the fabrication technique, size, processed and

their possible applications. There has been a dynamic range of biomedical applications of strain sensing including the breath monitoring (Ryu et al. 2015; Yamada et al. 2011), blood flow (Bingger et al. 2012), knee movements (Amjadi et al. 2015) and heart rate (Park et al. 2014). Even though the mercury-based strain gauges have been used for strain-sensing purposes for decades, some of the demerits of these sensors are the range of strain and the involvement of mercury for biomedical applications (Cai et al. 2013). Addressing these drawbacks, researchers have been trying to exploit different conductive materials like CNTs (Yamada et al. 2011; Yin et al. 2011; Zhao et al. 2010) and graphene (Jing et al. 2013; Qin et al. 2015; Wang et al. 2011) to develop strain sensors with higher sensitivity and efficiency in terms of conductivity, gauge factor, and potential applications. Although some of these sensors have been able to generate high sensitivities, their technical complications (Cai et al. 2013), high fabrication cost (Bae et al. 2013), and the requirement of experts to develop, operate and replace them has caused their limitations regarding their widespread uses. It is thus, state of the art to fabricate low-cost, efficient strain sensors that capable of rectifying the above-mentioned limitations. The developed sensors have been used to analyze the physiological bending movements.

7.4.1 Experimental Setup

The sensor patches were connected on different body parts like a finger, knee, elbow, and neck. The sensor patches were positioned in such a way that the bending of the joints induced strain on the sensing area of the patches. The metacarpophalangeal, olecranon and patella were the joints on the finger, elbow, and knee where the force was induced on the sensing area of the patches. The sensor patches were placed on the upper part of the cervical spine to determine the movement of the neck. The movements of the body parts were done in oscillatory motions to ensure the repeatability of the responses of the sensor patches and validate their functionality for strain-sensing purposes. The direction of movements for these joints was done in the vertical direction to induce strain in a perpendicular direction to the electrode fingers. These four joints were chosen for the strain sensing purposes as they exerted the highest force in those respective body

parts. The strain induced by these joints caused a change in the resultant electric field between the electrodes, thus changing the resultant output of the sensor patch. The analysis was done using an impedance analyzer via an automatic data acquisition algorithm.

7.4.2 Results and Discussion

A HIOKI IM 3536 High Precision Tester was used to connect the bonding pads of the sensor patch via Kelvin probes for recording the movement of the sensor patches. The analyzer was attached via a USB cable to store the data in the computer via Microsoft Office Suite. An alternating voltage of 1 V RMS was provided as the input signal where the frequency fixed at 5 kHz. An average of three readings with a duration of 5 seconds between each sweep was considered to validate the repeatability of the responses of the sensor patches. Figures 7.16 – 7.19 show the attachment of the sensor patches on different body parts and their respective responses. Figures 7.16 (a) and 7.16 (b) show the positioning of the sensor on the middle finger of a finger and its corresponding response respectively.

(a)

(b)

Figure 0.16 Attachment of the sensor patch on the (a) middle finger of the hand with the assistance of a bandage to determine its (b) movement based on the strain induced on the sensing area of the patches. The two states, flexed and extended, refer to the bent and relaxed condition of the arm respectively.

The middle finger was considered for monitoring purposes due to its position and its rate of movement in hand in comparison to the other fingers. The sensor patch was positioned on the finger via its attachment to the glove with super glue to ensure its proper attachment. The two states depicted in the graph, flexed and extended is related to the bent and relaxed conditions respectively of the finger. Figures 7.17 (a) and 7.17 (b) depict the attachment of the sensor patch on the elbow of the arm. The sensor patch was fixed on the elbow via a bandage. The use of the bandage to attached the sensor patch was considered to consider the potential of the sensors to be used for different applications in detection purposes like human-motion (Yamada et al. 2011) and physiological parameters (Yeo et al. 2013). The sensor patch was able to respond to the strain induced by the movement caused by the elbow and the finger. However, certain differences pertaining to the range of the conductivity values between the two alternate responses. The reason for this difference could be

due to the higher degree of movement of the elbow with respect to finger.

(a)

(b)

Figure 0.17 Attachment of the sensor patch on the (a) elbow of the arm with the assistance of a bandage to record its (b) movement based on the strain induced on the sensing area of the patches. The two states, flexed and extended, refer to the bent and relaxed condition of the arm respectively.

Figure 0.18 Attachment of the sensor patch on the (c) neck with the assistance of a bandage to record its (b) movements based on the strain induced on the sensing area of the patches. The two states, flexed and extended, refer to the bent and relaxed condition of the arm respectively.

(a)

(b)

Figure 0.19 Attachment of the sensor patch on the (a) knee to record its movements based on the strain induced on the sensing area of the patches. (b) The two states, flexed and extended, refer to the bent and relaxed condition of the arm respectively.

The attachment of the sensor patch was also done as shown in Figure 7.18 (a) to determine its movement. The idea for monitoring the neck movement is to induce the potentiality of the sensor towards people suffering from neck injuries or paraplegic patients who have problems of tweaking facial muscles (Rakibet et al. 2014). It is seen from Figure 7.18 (b) that the sensor patch was capable of detecting the oscillatory movement of the neck. The movement of the leg was determined as seen from Figures 7.19 (a) and 7.19 (b) by attaching the bandage-assisted sensor patch positioned on the patella of the left leg. The monitoring of the leg movement induces the potential of the sensors to detect the movements of the people having osteoporotic fractures. However, the range in the conductivity values obtained during flexed states is not the same for all the oscillatory cycles. There are two possible reasons for this behavior. Firstly, the positioning of the bandage assisted sensor patch on the joints might not be in the same manner for any two flexed cycles. The oscillatory motion of the leg might have led to a change in the orientation of the bandage from its preceding position. This aberrance will be minimized in the contexts where the monitoring would be done by fixing the sensor patch connected to an embedded system. Secondly, there might be minuscule changes in the physiochemical composition of the graphite powder in the nanocomposite formed electrodes in every cycle. This changes a change in the response of the sensor patch. Even though this change might be a concern regarding the repeatability of the responses of the sensor patches, the results would fluctuate within a certain range as a result of which, the optimization of the amount of graphite powder can be done in the resultant nanocomposite.

7.4.3 Conclusion

The sensor patches were considered for biomedical applications where they were fixed to the joints of the fingers and limbs to analyze their movements. The merits of using these sensors are their excellent electrical and mechanical properties leading to a higher probability in these types of applications. The sensitivity of the sensor patches was as per their responses to small and strains exerted from a finger and limbs respectively. There is however one issue that the sensor patches faced during the experiments, which need to be rectified for further

tuning of the responses of the sensor patches. The experiments were performed using an impedance analyzer which was with wired connections. This could be avoided in real-time applications where wireless communications like active Radio-frequency Identification (RFID) tags can be added on the sensor patches so that they could transmit the signal to the reader at fixed intervals. Also, instead of attaching the sensor patches with the biocompatible tapes, they can also be cuffed for ubiquitous monitoring purposes.

7.5 Force Sensing

The sensor patches were also employed for low-force sensing purposes where the patches measured forces ranging between 3.5 mN to 17.5 mN. Low-force sensing holds a pivotal role in certain applications like biomedical measurements, microfluidics and tactile sensing (Wong et al. 2012). In certain cases like tactile sensing, they have been employed for robotic and upper limb prostheses (Tiwana et al. 2011), in touch screens which fabricated using tactile sensors (Kim et al. 2011a), and soft tactile sensors were achieved using it (Viry et al. 2014). Although a range of materials has been processed in these research works, there have been certain limitations like the high cost of processing, the undetectable range of operation, complexity in the design and operating principle and rigidity, that are associated with them causing a limitation to their applications. For example, even though the microfluidic multilayer sensor can sense forces up to 2.5 N (Wong et al. 2012), their disadvantages lie in the cost of fabrication and complexity of the structure. Although the research from (Lee et al. 2011) have worked on capacitive polymer tactile sensors, the detectable range, in this case, is too high (20 mN), especially in case of haptic robotic applications. Our work provides a conjunctive approach deterring some of the attributes like the cost of fabrication, operating principle, and detectable range. Our sensors are also able to detect forces at around 3.5 mN, which increases their potential as wearables for rehabilitation purposes after the patient suffers from a stroke and muscle spasms. In such cases, the sensors would be able to monitor of minuscule movements of a body part to determine the patient's recovery.

7.5.1 Experimental Setup

The experimentation of the sensor patches was done using an impedance analyzer (HIOKI IM 3536 High tester) connected to the sensors using Kelvin probes from one end. The sensors were attached to the clamps or bent during the experimental procedure. The experiments were carried out by firmly fixing the sensor patches at the end opposite to Kelvin probe to analyze the changes due to the applied load.

7.5.2 Results and Discussion

The testing on the sensor patches was done using five different forces, ranging between 3.5 mN and 17.5 mN. The patches were firmly fixed to minimize any movements during its experimentation. Different weights of the same shape were placed on the same location in the patch to avoid the effects of their area on the results.

Figure 0.20 Response of the sensor patch for a particular frequency (5 kHz) depicting the different capacitance values for the different experimental weights.

An average of five experimental readings was taken for each weight to validate the repeatability of the responses. The experiments

were done at a frequency value of 5 kHz to analyze the changes in capacitance values as a function of the loading force. As shown in Figure 7.20, the change in capacitance with respect to force is almost linear obtaining a force sensitivity of $0.2542 \text{ pF.mN}^{-1}$ for the tested range.

7.5.3 Conclusion

Followed by the biomedical applications for measuring the physiological movements, the fabricated Graphite/PDMS sensor patches were also used for low-force sensing applications. The sensors obtained a force sensitivity of $0.2542 \text{ pF. mN}^{-1}$ in the test range between 3.5 mN and 17.5 mN. The results show the capability for the developed sensor patches to be used as low-cost and highly efficient force sensors, which can be used to form efficient force-sensing systems for different ubiquitous monitoring applications.

7.6 Chapter Summary

The fabrication, characterization and application of the novel Graphite/PDMS sensor patches were showcased in this chapter. The novelty in this approach was based on the development of the simplified sensor patches done using casting technique based on 3D printed molds. The casting of graphite and PDMS were done on the trenches of the moulds and above it respectively to form the electrode and substrate parts of the patches, respectively. The merits of these patches lie in its low-cost fabrication, simple operating principle and high performance for the designated applications. Another major merit is the reliability of the sensor's fabrication on a large scale using this technique in terms of their structure and repeatability in the response during characterization and experimentation. Since the 3D printed molds were used as the main template to cast graphite and PDMS on them; it is possible to develop the sensor prototypes with identical dimensions having similar performances. Once the design of the sensor structure was done, the fabrication of all the 3D printed molds took very less time (around 30 mins including the pre-heating process). After the optimization of the amount of graphite powder to be used in the sensor patch, identical performance of the electrode

with respect to the electrical conductivity was achieved. The casting of the PDMS was followed by of its height was also being able to be done identically for all the sensor patches. The differences between the performances of the sensor prototypes in terms of its characterization and experimentation were negligible. The responses of the sensor patches for their characterization and experimentation were analyzed in terms of the change in the electrical conductivity and capacitance with the induced strain. Due to the flexible nature of the fabricated sensor patches, they were capable of responding to the stress-strain experiments and oscillatory bending cycles. The use of the 3D printing technique ensured the low-cost of the raw material to form the molds, low fabrication time and high repeatability with respect to the structural dimensions of the developed sensor patches. The amalgamated attributes of PDMS and graphite in terms of their electrical and mechanical properties ensured the flexibility of the sensor patches and high electrical conductivity of the electrodes. Other than the showcased applications, these sensor patches have the potential to be employed for other biomedical uses like monitoring of blood pressure and oxygen saturation (Bingger et al. 2012), as a temperature sensor (Shih et al. 2010), determining the bone density and mechanical properties of bones (Bhalla and Bajaj 2008; Dharap et al. 2004), soft robotics (Jeong et al. 2015; Kang et al. 2014), piezo-resistive sensing (Zhao et al. 2010), virtual reality and for entertainment purposes (Amjadi et al. 2014).

7.7 References

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