# Multilayer Capacitive Pressure Sensor

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**ABSTRACT** This work introduces a novel geometric design for multilayered structures, enhancing resistance to delamination and reducing surface stress and strain during bending deformation. Employing polydimethylsiloxane (PDMS) as a low-modulus material and polyethylene terephthalate (PET) as a high-modulus material, the study explores the fabrication and performance of capacitive sensors. PDMS is favored for its flexibility, biocompatibility, and ease of fabrication, making it suitable for various sensing applications, particularly in biomedicine. PET, on the other hand, boasts excellent dielectric properties, mechanical flexibility, and dimensional stability, making it ideal for creating robust and multifunctional flexible sensors. The fabrication process involves meticulous steps, including mold design, substrate preparation, sensor assembly, and curing, aiming to achieve precise geometries and reliable performance. Experimental results demonstrate the sensor's calibration, sensitivity, repeatability, and reproducibility. Failed attempts highlight challenges in sensor fabrication, such as insensitivity to pressure and the presence of bubbles between PDMS and PET layers. The study concludes with insights into sensor performance and future research directions to address fabrication challenges and enhance sensor capabilities for diverse applications.

**INDEX TERMS** Capacitive pressure sensors, Flexible pressure sensors, Biomedical applications, Wearable devices, Sensor fabrication, Polydimethylsiloxane (PDMS),Polyethylene terephthalate (PET), Multilayer structures, Geometric design, Delamination resistance, Surface stress, Strain, Calibration, Sensitivity, Repeatability, Reproducibility, Mold design, Curing process, Optimization

## I. INTRODUCTION

With decreased power consumption, less sensitivity to temperature fluctuations, and a lower fundamental noise level, capacitive pressure sensors are increasingly replacing piezoresistive ones in the market. There are several benefits of employing capacitive pressure sensors. They are appropriate for battery-operated devices and applications where power efficiency is crucial since, among other things, they have low energy consumption. Second, great repeatability provided by capacitive sensors guarantees accurate and consistent readings over an extended period of time. Furthermore, they remain sensitive throughout environmental fluctuations, offering accuracy and stability throughout a range of operating circumstances. Usually, a thin, flexible dielectric layer sits between two parallel plates to form a capacitive pressure sensor. While the top plate is flexible and moves in reaction to pressure, the bottom plate is immovable. The variable capacitor is formed by the space between the plates. Applying pressure causes the dielectric layer to deflect, which modifies the capacitance by changing the spacing between the plates. The applied pressure and the change in capacitance are exactly proportional. [1]

The technology known as Electronic Skin (e-skin) has revolutionized the way humans and machines interact. E-skin with built-in pressure sensors mimics the suppleness and responsiveness of human skin, allowing for accurate and natural contact with electronic devices. Flexible pressure sensors have a lot of potential for use in medical applications, thus much research has gone into making them more sensitive. Biomedical applications require wearable and highly flexible devices. As a result, the devices' sensitivity and size have been limited. Since these devices are meant to come into touch with human skin and undergo constant deformation and bending in their shape, they should have a high enough level of tolerance power to survive long-term mechanical deformation such stress, bending, and stretching. The materials we utilize to fabricate these sensors go through the same mechanical changes that have an impact on the sensor's overall lifespan and output. The device's design and material selection are extremely important in this situation. For flexible and bendable device applications, it is therefore necessary to verify and optimize the various layer geometries in order to increase device reliability. The advantages of changing

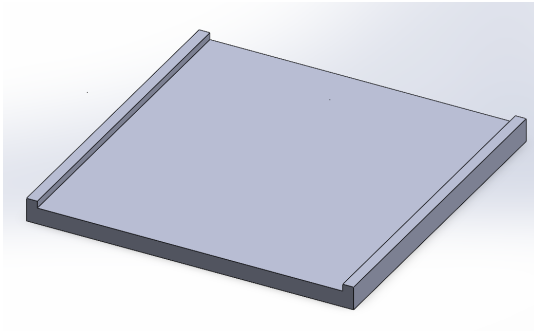

FIGURE 1. Rectangular Cross-Section

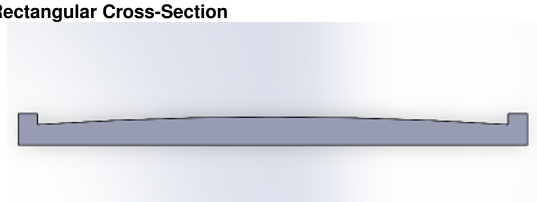

FIGURE 2. Concave Cross-Section

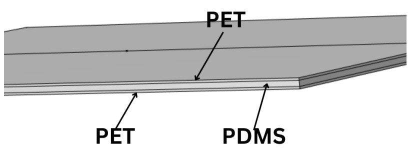


FIGURE 3. 3D design of Rectangular Sensor

the geometry of the layers in a multilayer construction are further supported by the way the sensor properties behave on top of the suggested substrate during bending cycles in this work. To lessen surface strain on the outer layers—which are frequently where electronics sensors and circuits are installed—multilayer substrates contain a sandwich construction with soft material, or low modulus material, in the middle and hard material, or high modulus material, on the outside. [2]

In this work, we offer a novel geometrical design for the multilayered structure, which is more resistant to delamination than the standard multilayered one and exhibits less generated surface stress and strain during bending deformation. Here, we employed polydimethylsiloxane (PDMS) as a low-modulus material and polyethylene terephthalate (PET) as a high-modulus material.

## II. Material Description

### A. polydimethylsiloxane(PDMS)

In the field of sensor construction, polydimethylsiloxane (PDMS) is a fundamental material that is highly valued for a distinct set of characteristics that render it perfect for a broad range of sensing applications. Because of its exceptional flexibility, biocompatibility, and optical transparency as an elastomer based on silicone, researchers and engineers working on the development of novel sensing devices in a variety of fields—such as consumer electronics, biomedicine, and environmental monitoring—preferred to work with PDMS.
Fundamentally, PDMS has a unique molecular structure made up of repeating dimethylsiloxane (-Si(CH)O-) units connected by siloxane bonds, which gives it remarkable mechanical capabilities. Because of its low modulus and high elongation at break, PDMS is a very flexible and deformable material that can be used to create soft, conformable sensors that can easily interface with a variety of surfaces, including biological tissues.

The biocompatibility of PDMS, which makes it ideal for biomedical sensing applications, is one of its most alluring features. Because of its low cytotoxicity and tissue reactivity, PDMS is compatible with biological systems and makes it easier to construct wearable health monitors, lab-on-a-chip devices, and implantable sensors for diagnostic and therapeutic applications.

The fact that PDMS may be used with several fabrication methods, including as microfluidics, replica molding, and soft lithography, adds to its attractiveness for sensor creation. By using these methods, researchers can produce microfluidic channels, integrated electrode arrays, and complicated sensor geometries with great repeatability and precision, opening the door to the realization of sophisticated sensing architectures that are suited to particular application needs. [3] [4] [5] [6]

### B. PET

Polyethylene terephthalate (PET) appears as a promising material in the field of flexible capacitive sensor fabrication, with a unique set of features that allow for the creation of robust and multifunctional sensing devices. In this section, we look at how PET is used as a foundational material in the production of flexible capacitive sensors, focusing on its features, processing methods, and applications in this specialized field.

PET has characteristics that make it a good choice for applications involving capacitive sensors. PET is a great insulator for generating the dielectric layer of capacitive sensors because of its inherent dielectric qualities, which include low dielectric loss and high dielectric strength. Furthermore, even under changing climatic conditions and mechanical loads, PET's mechanical flexibility, resilience, and dimensional stability guarantee the lifespan and dependability of flexible sensor structures.

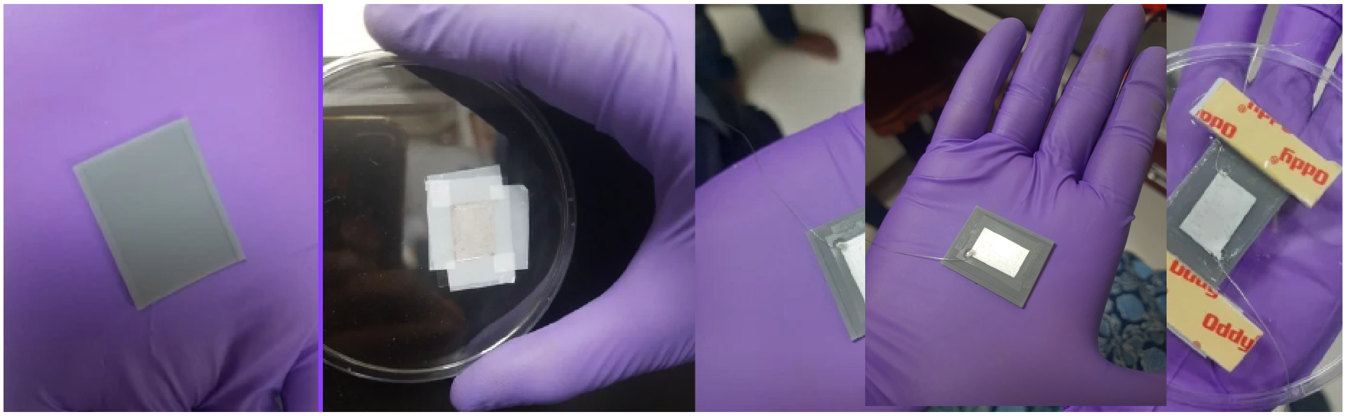



**FIGURE 4. Fabrication Steps shown are: Mold, Masking, Pouring PDMS, Curing**

PET-fabricated flexible capacitive sensors are used in a wide range of industries, such as biomedical devices, wearable electronics, and human-machine interfaces. PET-based capacitive sensors in wearable electronics can be included into clothes, textiles, or accessories to track fitness and monitor physiological factors like breathing, muscular activity, and body movements. PET-based sensors improve user interaction and experience in human-machine interfaces by enabling touch-sensitive surfaces and gesture recognition features in flexible displays, touchscreens, and interactive gadgets. PET-based capacitive sensors can be used in biomedical devices to detect biological signals non-invasively, including electrocardiography (ECG), pulse oximetry, and heart rate. This allows for remote patient monitoring and diagnostic applications. [9] [10] [11] [12]

### C. Silver as a electrode

Within the field of capacitive sensor manufacturing, silver painting is a flexible and economical technique for producing electrode layers with accurate geometries and superior electrical conductivity. This section examines the use of silver painting as an electrode layer in the manufacture of capacitive sensors, as well as its benefits and manufacturing process.

Because of its excellent electrical conductivity, silver is a perfect material for the electrode layers that are formed in capacitive sensors. Silver's low resistivity makes for effective signal detection and charge transfer, which raises the sensor's sensitivity and performance.

Strong adherence is exhibited by silver paints or inks on a range of substrates, including silicon, glass, polymers, and ceramics, which are frequently employed in the manufacture of sensors. This adhesion minimizes signal loss and increases the sensor's lifetime by ensuring a stable contact between the electrode layer and the substrate.

There is versatility in electrode design and patterning when it comes to silver painting. Precise electrode shapes, including as spiral electrodes, comb electrodes, and interdigitated electrodes, can be deposited according to the application's particular sensing needs. [13]

## III. Experimental work

The mold is the base where sensor has been fabricated. According to the need of device, mold has to be printed in different size and dimension.

### A. Mold Designing and CAD work

For our desired sensor design, $3cm \times 3cm$ mold has been designed in Solidworks. Thickness and boundary height of the mold is 1.5mm. The first part of the mold is fabricating in sensor when substrate is flat as shown in fig 1

The second design is for fabricating the trilayer sensor when the substrates are in concave shape.

### B. Fabrication of the Sensor

This method shows the fabrication of rectangular trilayers sensor. The aim is to fabricate the sensor with dimension of $1.5cm \times 2cm$, PET of thickness 50 micron each and PDMS of thickkness 200 micron. The following steps have been followed for the fabrication:

- Take PDMS and crosslinker in the ratio of 20:1. We took 12.5g PDMS and 2.5g crosslinker.
- Mix PDMS and crosslinker together and stirr them well till 10-15 minutes.
- Put the mixture into fridge for few hours to get rid of bubbles
- Cover the open side of the mold with double sided tape.
- Take PET and Tapped it on a base then do silver coating of dimension $1.5cm \times 2cm$.
- Attach the wires with epoxy on the silver coated PET.
- Paste a silver coated PET with double sided tape on the mold.
- Slowly slowly pour PDMS on the mold now and avoid the formation of bubbles.
- Put the sample into oven at 70 degree Celcius for 10 minutes.
- Take out the sample and stick the another silver coated PET on the top. Take care of the air bubbles while sticking it to the top surface of PDMS.

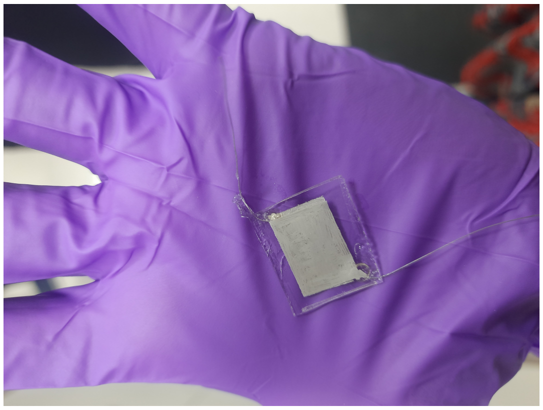

FIGURE 5. **Fabricated cured capacitive sensor**

- Put back the sample in oven for 5-6 hours at the same temperature.
- Take out the fabricated sensor from the mold.

### *C. Other Fabrication*

Apart from the above we have fabricated the sensors following the previous methods for the different compositions of the substrate. In the fisrt case we changed the ratio of PDMS and crosslinker and took it to 25:1. The PDMS was 8.3g and crosslinker was 0.33g and fabricated the sensor using the same method.
In the second case we took the same ratio of PDMS and crosslinker as 25:1 but PET thickness of 25 micron rather than 50 micron.

### *D. Concave Cross-section Fabrication*

In this type of fabrication,we introduce a geometrical design for the multilayered structure which shows less developed surface stress and strain under bending deformation and is also more immune to delamination than the conventional multilayered one as shown in Fig.6. For fabcricating this, we have used the similar fabrication method on the mold shown in Fig.2. After final curing process, we get our sensor as shown in Fig.7.

### *E. Testing Setup*

For the characterization and sensitivity analysis of each sensor, we employed a setup depicted in Figure 10. The sensor was interfaced with an LCR meter. To commence the process, we specified the dimensions and area of the sensor and selected the appropriate capacitance (Cp) for subsequent calculations. Subsequently, a frequency sweep was conducted on the computer prior to obtaining final readings. The readings were then initiated, wherein the weight applied to the sensor was incrementally or decrementally adjusted in a gradual manner.

In Figure 9, the experimental setup for sensor characterization was illustrated. This involved placing the sensor on a human hand and measuring the pressure exerted by the hand pulse.

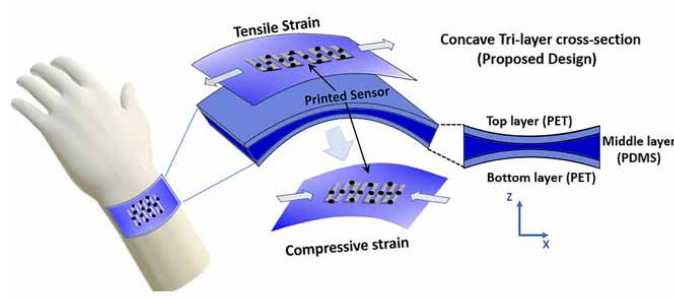


FIGURE 6. **Concave structure Sensor [1]**

FIGURE 7. **Cross-sectional View of fabricated capacitve sensor**

## IV. Result And Discussion

### *A. Calibrating the sensor*

In the Fig.8 we can see the capacitance vs pressure plot that represents the input-ouput characteristic of the capacitance pressure sensor. We can observe that capacitance increases with the applied pressure that satisfy the theoretical condition of a capacitor as capacitance is inversely proportional to the gap between the two capacitor plates. When we are applying the pressure here then there is deformation in PDMS and it is pressed down hence distance between two silver coated PET decreases that implies increment in capacitance value.
From the plot it is clear that the sensor is following a logarithmic characteristic. After calibrating the sensor from the observed data, we got the following calibration equation:

$$C_p = 0.38 \times logP + 6.985 \tag{1}$$

The blue curve represents the characteristic of the sensor when operated in forward direction with increasing load value. Orange curve represents the hysteresis of the sensor that seems to be quite low.

### *B. Sensitivity*

We know that sensitivity is the slope of the calibration curve at any point of time. We can simple obtain it by

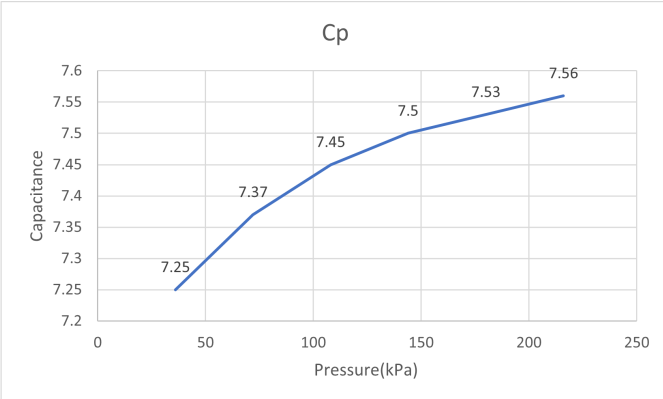


**FIGURE 8.** **Calibration Plot**

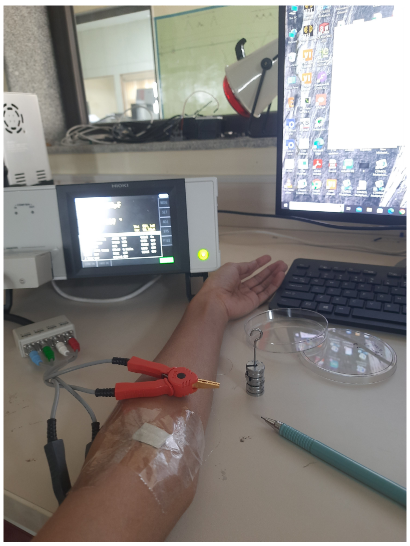

**FIGURE 9.** **Testing for pulse pressure**

differentiating capacitance with respect to pressure at any point of time

$$\frac{dC_p}{dP} = \frac{0.38}{ln10 \times P} \tag{2}$$

### C. Repeatability and Reproducibility

Sensor ran only once and after that it got dead hence these parameters are tasks to work on.

### D. Failed Attempts

The sensors we fabricated with ratio 25:1 and 50 micron PET were a failed attempts. It was not sensitive to pressure and sensitivity was almost zero.In the Fig.8 you can see that plot is almost flat and there is no change in capacitance.
Later we tried to fabricate with 25 micron PET but there were so many bubbles between the PDMS and PET layer. That is also need to be rectified.

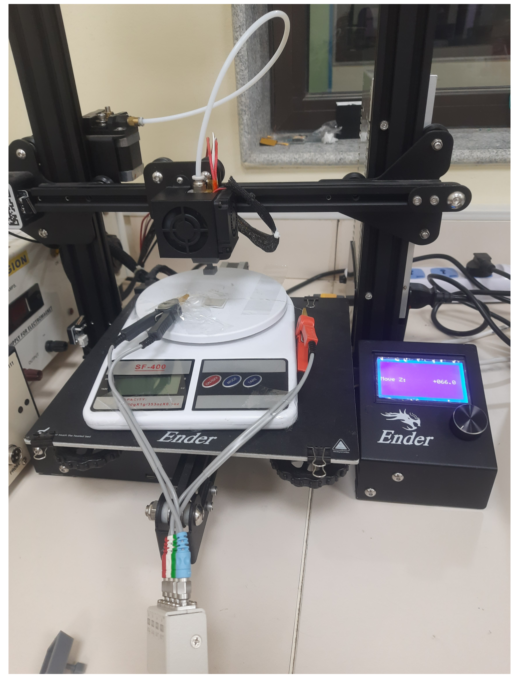


**FIGURE 10.** **Characterization Set-up**

### E. 25 Micron Fabricated Sensor Performance

In the earlier fabricated sensor, because of thick PET used there were not enough fexiblity for wearable application. Hence we fabricated the sensor using thin PET of 25 micron.

#### 1) Sensor No. 1

The sensor under investigation was fabricated employing a 1:30 ratio of cross-linker to polydimethylsiloxane (PDMS) utilizing a 25-micron polyethylene terephthalate (PET) substrate and total thickness of 1mm. Upon integration with an LCR (Inductance, Capacitance, Resistance) meter, an initial capacitance reading of 7.3 picofarads (pF) was observed, which subsequently increased proportionally with applied pressure, as depicted in the accompanying figure. The plateau region in the plot denotes the capacitance level achieved upon application of load to the sensor. The range of change is 0.3 from the initial capacitance value for both level of testing cycle.

A consistent correlation between capacitance and pressure was established, with capacitance demonstrating an escalating trend in tandem with increasing pressure. Upon subsequent testing iterations, a noticeable shift in the sensor's baseline capacitance was observed, accompanied by the manifestation of hysteresis. This phenomenon is likely attributable to the physical stress exerted upon the sensor during operation.

#### 2) Sensor No. 2

Figure 13 illustrates the operational characteristics of sensor two, which was produced utilizing the same fabrication

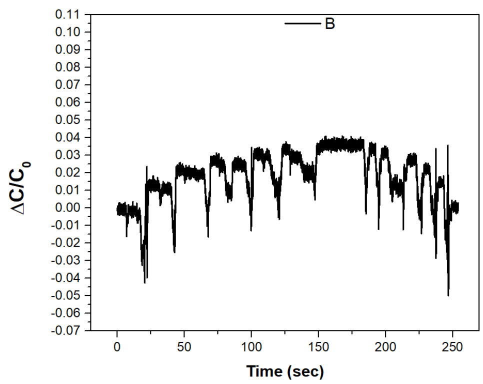


FIGURE 11. Sensor 1: Change in capacitance with applied pressure

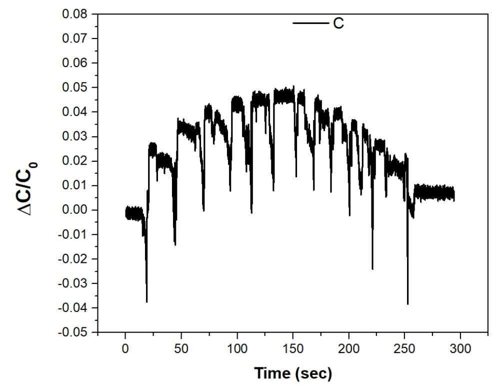


FIGURE 13. Sensor 2: Change in capacitance with applied pressure

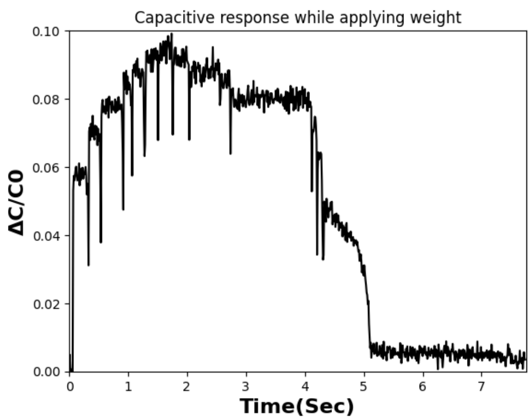


FIGURE 12. Sensor 1: Change in capacitance with applied pressure(cycle-2)

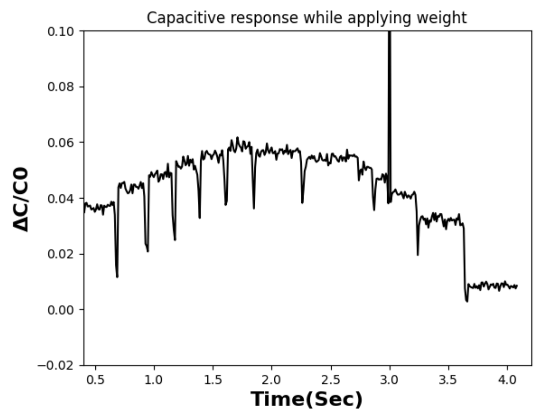


FIGURE 14. Sensor 1: Change in capacitance with applied pressure(cycle-2)

technique with a total thickness of 800 microns. The sensor exhibited an initial capacitance reading of 8.5 picofarads (pF), which increased to 9.00 pF under a maximum applied pressure of 216 kilopascals (kP). Notably, significant hysteresis was observed in Figures 13 and 14; however, its magnitude was comparatively lower than that observed in sensor one. Additionally, sensor two demonstrated an enhanced performance range compared to sensor one, rendering it particularly suitable for various biomedical wearable applications.

### 3) Concave cross-section sensor

The concave sensor was fabricated using the same methodology as previously described, with a thickness of approximately 750 microns. Upon characterization, the initial capacitance recorded was 16.3 picofarads (pF), which increased to a maximum of 16.8 pF under a pressure of 216 kilopascals (kP). It was noted that all sensors exhibited a consistent range of incremental capacitance, typically between 0.3 to 0.5 pF per unit of applied pressure. Notably, the concave sensor demonstrated a broader range of capacitance variation, with an increment of 0.5 pF, compared to the standard sensor's increment of 0.3 pF.

Furthermore, the concave design exhibited superior mechanical strength and stability, coupled with exceptional flexibility and resistance to delamination. These qualities signify the potential for enhanced performance and longevity in practical applications.

### 4) Testing Sensor on Hand

In Figure 9, the experimental setup utilized for sensor characterization is depicted. Meanwhile, in Figure 16, a continuous increment in capacitance is observed in response to variations in heart pulse pressure. In Figure 9, the experimental setup utilized for sensor characterization is depicted. Meanwhile, in Figure 16, a continuous increment in capacitance is observed in response to variations in heart pulse pressure.

The decreasing portion in each of the sensors mentioned above is deloading part where pressure has been decrement back to 36 kPa for studying the hysteresis

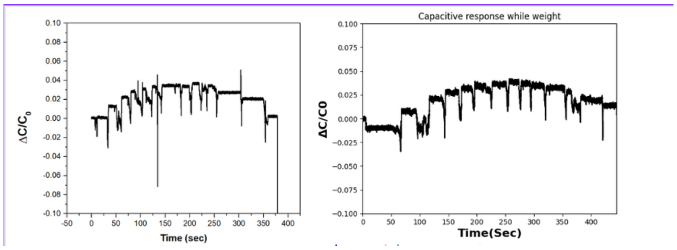


**FIGURE 15.** **Concave Sensors: Change in capacitance with applied pressure**

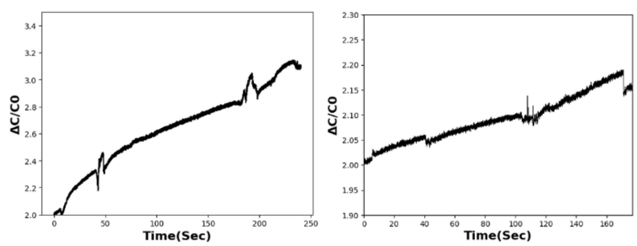


**FIGURE 16.** **Testing on hand: Change in capacitance with pulse rate**

## V. Conclusion

In conclusion, capacitive pressure sensors have emerged as a promising alternative to piezoresistive ones, offering advantages such as reduced power consumption, enhanced stability in varying temperatures, and lower fundamental noise levels. The study highlights the significance of capacitive sensors in diverse applications, particularly in biomedical and wearable devices, where power efficiency and accuracy are paramount.

The integration of capacitive sensors into Electronic Skin (e-skin) has revolutionized human-machine interactions, enabling natural and precise contact with electronic devices. Flexible pressure sensors, a key component of e-skin, hold immense potential in medical applications, driving ongoing research to enhance their sensitivity and durability.

Fabrication of capacitive sensors involves careful selection of materials and optimization of multilayer structures to ensure resistance to delamination and minimize surface stress during bending deformation. Polydimethylsiloxane (PDMS) and polyethylene terephthalate (PET) have emerged as promising materials due to their flexibility, biocompatibility, and dielectric properties.

Experimental results demonstrate the calibration, sensitivity, repeatability, and reproducibility of fabricated sensors. Challenges such as insensitivity to pressure and bubble formation between PDMS and PET layers underscore the need for continued research to address fabrication complexities and enhance sensor performance.

In future research, efforts should focus on refining fabrication techniques, optimizing material properties, and exploring novel sensor designs to overcome existing challenges and unlock the full potential of capacitive pressure sensors in various applications. With ongoing advancements in sensor technology and interdisciplinary collaboration, capacitive sensors are poised to play a pivotal role in shaping the future of wearable devices and biomedical technologies.

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## REFERENCES

[1] K. Tripathy and M. Bhattacharjee, "Mechanical Neutral Plane Modulation in Polymer Flexible Multilayer Substrate for Ultrastable Sensor Fabrication," in IEEE Sensors Journal, vol. 23, no. 20, pp. 24231-24238, 15 Oct.15, 2023, doi: 10.1109/JSEN.2023.3310342. keywords: Strain;Substrates;Bending;Sensors;Nonhomogeneous media;Stress;Deformation;Bending deformation;flexible sensing platform;multilayer substrate;neutral mechanical plane;wearable sensor,

[2] Javidi, R., Moghimi Zand, M. Alizadeh Majd, S. Designing wearable capacitive pressure sensors with arrangement of porous pyramidal microstructures. Micro and Nano Syst Lett 11, 13 (2023). https://doi.org/10.1186/s40486-023-00178-7

[3] Whitesides, G. M., Xia, Y. (1998). Soft Lithography. Angewandte Chemie International Edition, 37(5), 550-575. [DOI: 10.1002/(SICI)1521-3773(19980316)37:5¡550::AID-ANIE550¿3.0.CO;2-G]

[4] McDonald, J. C., Duffy, D. C., Anderson, J. R., Chiu, D. T., Wu, H., Schueller, O. J., Whitesides, G. M. (2000). Fabrication of microfluidic systems in poly (dimethylsiloxane). Electrophoresis, 21(1), 27-40. [DOI: 10.1002/(SICI)1522-2683(20000101)21:1¡27::AID-ELPS27¿3.0.CO;2-C]

[5] Sia, S. K., Whitesides, G. M. (2003). Microfluidic devices fabricated in poly (dimethylsiloxane) for biological studies. Electrophoresis, 24(21), 3563-3576. [DOI: 10.1002/elps.200305584]

[6] Qin, D., Xia, Y., Whitesides, G. M. (2010). Soft lithography for micro- and nanoscale patterning. Nature Protocols, 5(3), 491-502. [DOI: 10.1038/nprot.2009.234]

[7] Kim, D. H., Lu, N., Ma, R., Kim, Y. S., Kim, R. H., Wang, S., ... Rogers, J. A. (2011). Epidermal electronics. Science, 333(6044), 838-843. [DOI: 10.1126/science.1206157]

[8] Zhao, X., Kim, J., Cezar, C. A., Huebsch, N., Lee, K., Bouhadir, K., ... Mooney, D. J. (2011). Active scaffolds for on-demand drug and cell delivery. Proceedings of the National Academy of Sciences, 108(1), 67-72. [DOI: 10.1073/pnas.1012864108]

[9] Kim, D. H., Lu, N., Ma, R., Kim, Y. S., Kim, R. H., Wang, S., ... Rogers, J. A. (2011). Epidermal electronics. Science, 333(6044), 838-843. [DOI: 10.1126/science.1206157]

[10] Kwon, H. J., Lee, K. J., Yi, G. C., Lee, S. W., Kim, S. H. (2016). Flexible capacitive sensors with patterned polymer dielectric layers for human machine interfaces. Scientific reports, 6, 24486. [DOI: 10.1038/srep24486]

[11] Mannsfeld, S. C., Tee, B. C., Stoltenberg, R. M., Chen, C. V., Barman, S., Muir, B. V., ... Bao, Z. (2010). Highly sensitive flexible pressure sensors with microstructured rubber dielectric layers. Nature materials, 9(10), 859-864. [DOI: 10.1038/nmat2834]

[12] Mannsfeld, S. C., Tee, B. C., Stoltenberg, R. M., Chen, C. V., Barman, S., Muir, B. V., ... Bao, Z. (2010). Highly sensitive flexible pressure

sensors with microstructured rubber dielectric layers. Nature materials, 9(10), 859-864. [DOI: 10.1038/nmat2834]

[13] Ma, Zhuyu Zhang, Yang Zhang, Kaiyi Deng, Hua Fu, Qiang. (2022). Recent progress in flexible capacitive sensors: Structures and properties. Nano Materials Science. 10.1016/j.nanoms.2021.11.002.

[14] Zhao, X., Kim, J., Cezar, C. A., Huebsch, N., Lee, K., Bouhadir, K., ... Mooney, D. J. (2011). Active scaffolds for on-demand drug and cell delivery. Proceedings of the National Academy of Sciences, 108(1), 67-72. [DOI: 10.1073/pnas.1012864108]

[15] Someya, T., Kato, Y., Sekitani, T., Iba, S., Noguchi, Y., Murase, Y., ... Sakurai, T. (2005). Conformable, flexible, large-area networks of pressure and thermal sensors with organic transistor active matrixes. Proceedings of the National Academy of Sciences, 102(35), 12321-12325. [DOI: 10.1073/pnas.0501391102]