A HIGHLY SENSITIVE MODULAR ELASTOMER-BASED CAPACITIVE PRESSURE SENSOR FOR RESPIRATION MONITORING

by

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A Highly Sensitive Modular Design For Elastomer-Based Capacitive Pressure Sensor For Respiration Monitoring

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Abstract

Sleep Apnea (SA) is one of the most prevalent but yet underreported public health issues all around the world. The diagnosis of SA requires sleep monitoring or polysomnography (PSG), which is recording several bio-physiological signals during sleep, including respiration. The state of the art for respiration monitoring is measuring the nasal or oral airflow by means of a pressure sensor, placed on the chest of the patient and connected to a long nasal/oral cannula (tube) that is placed in front the nose/mouth. This long tube in many cases is a source of discomfort for the patient. In this work, an elastomer-based flexible capacitive pressure sensor is presented that mounts on the upper lip of the patient. The sensor structure has a novel modular design that gives freedom to make the sensor smaller or larger with the same relative sensitivity. The sensor is designed for the low pressure range from -50 to 100 Pa suitable for the intended application with a high absolute and relative sensitivity of 0.52 pF/Pa and 2.064 kPa⁻¹, respectively, and capability to withstand overpressure up to 2 kPa without permanent damage. The sensor prototype has low non-linearity and hysteresis errors of less than 12%FS a resolution of 1.7 Pa.
Lay Summary

The focus of this work is the development of a more comfortable and reliable respiration monitoring device for sleep apnea diagnosis compared to current technology. Sleep apnea is a prevalent sleep disorder among people worldwide, that causes the patient’s breathing repeatedly to stop during sleep. It causes sleepiness during the day, snoring, and in the long term, a high chance of heart attacks. Current technology for respiration monitoring uses a long tube that goes from the nostrils to a pressure sensor placed on the patient’s chest and creates discomfort during sleep and also may dislodge and invalidate the sleep study; in contrast, our proposed flexible device will be directly mounted on the upper lip and does not require tubes. It enables inexpensive sleep monitoring in every household to support early diagnosis of sleep apnea.
Preface

This thesis is an original, unpublished work by the author, Hamed Pouriyayevlai, at the University of British Columbia (UBC) under the supervision of Prof. Boris Stoeber, Professor of Mechanical Engineering and Electrical and Computer Engineering, UBC. The research presented as “A Highly Sensitive Modular Design For Elastomer-Based Capacitive Pressure Sensor for Respiration Monitoring” was independently conducted by the author, as a sub-task of the comprehensive project “Sensory Information Technology and Implementation for Sleep Disorder Monitoring”, funded by the Natural Sciences and Engineering Research Council (NSERC) of Canada through the Strategic Partnership Grants project STPGP 493908.
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To my parents.
CHAPTER 1: INTRODUCTION

1.1 Sleep apnea

Sleep apnea is a sleep disorder that affects an estimated 20% of people worldwide [1]. A person suffering from sleep apnea experiences periodic interruptions to their breathing during sleep. These periodic interruptions can take the form of reduced ventilation (hypopnea), or a complete cessation of ventilation (apnea). Both apnea and hypopnea are considered to have similar impacts on the patient’s overall health [2]. These two types can be further divided into central sleep apnea (CSA) and obstructive sleep apnea (OSA). In CSA, the signals from the brain are not sent to the respiratory muscles. OSA is more common than CSA and is caused by the partial or complete collapse of the upper airway [2].

In 2017, an estimated 30 million adults in the United States had OSA [3]. Patients suffering from sleep apnea experience loud snoring, excessive daytime sleepiness [4], insomnia, nocturia, and morning headache. In some cases, patients will have sleep apnea and be asymptomatic [5]. For those who show the preliminary symptoms and who also possess certain risk factors for OSA, methods such as the Berlin questionnaire and the Epworth sleepiness scale can be used by the patient to diagnose themselves with OSA [6]. Still, many people with OSA remain undiagnosed because they are asymptomatic, or they show symptoms but fail to bring them to the attention of their healthcare providers.

1.2 Sleep apnea diagnosis

Diagnosing sleep apnea is accomplished by means of studying several bio-physiological signals during sleep, a process known as sleep monitoring or polysomnography (PSG). PSG is accomplished in one of four ways. Type I PSG is administered in a clinical setting, whereas Types II – IV PSG involve portable systems for home sleep apnea testing (HSAT). Type I PSG is generally considered the reference standard to which Types II – IV are compared [7].

For Type I PSG, the patient sleeps in a sleep clinic under continuous supervision of a sleep technician, and sensors are wired to the patient to record multiple channels of physiological signals such as brain activity (electroencephalogram, or EEG), muscle activity (electromyogram, or EMG) from chin and legs, eye movement (electrooculogram, or EOG), chest and abdominal movement (inertial measurement unit, or IMU), heart rhythm (electrocardiogram, or ECG), respiratory airflow from the nose and/or mouth,
and blood oxygen saturation level. However, Type I is inherently biased by the fact that a patient’s sleep quality in the laboratory setting is not representative of the sleep quality they would experience at home.

One way in which doctors can minimize the bias inherent to Type I PSG is to provide the patient with a portable setup to take home. The patient is given detailed instructions, PSG is performed unattended, overnight in their home, and the setup is brought back to the clinic the next day for analysis. As defined by the American Academy of Sleep Medicine (AASM), Type II must have a minimum of seven monitoring channels, Type III must have a minimum of four channels, and Type IV can have a single or two-channel measurement [8]. Respiration monitoring is always included in Types II-IV. While HSAT has an inherently lower diagnostic ability due to the limited number of channels, the results are more representative of what patients would experience in everyday life since they do not sleep as well in a laboratory setting as they would at home [9].

Studies show that obesity, aging, and male sex are the main risk factors of OSA [10], [11]. The cumulative number of apnea and hypopnea events per hour of sleep is known as the apnea-hypopnea index (AHI), and a patient with AHI over 5, 15, or 30 is considered to have mild, moderate, or severe OSA, respectively [12]. Prevalence of OSA in the general adult population (aged between 30 to 80 years old depending on the study) from several countries including India [13], Switzerland [14], Brazil [15], South Korea [16], for AHI ≥ 5 and AHI ≥ 15 ranges from 9% to 38%, and 6% to 17%, respectively. As mentioned earlier, males are more prone to OSA with a 13% to 33% compared to females with 6% to 19% [11].

Over the past 3 decades, obesity has increased among the population [17], [18] and more people are at higher risk of OSA [11], [19]. Many people do not take the OSA symptoms serious enough to discuss them with a health care center [12]. It is reported that only 3% of Canadian adults, 18 years and older, are reported to have been diagnosed with sleep apnea that 23% of the ones diagnosed with OSA were never referred to a sleep clinic to take an overnight test; also, 26% of Canadian adults are showing symptoms of having or developing OSA [20], and they could easily be tested with a more accessible way for examining their sleep disorders, such as a Type IV device. Considering the difference between Type I-measured and Type IV-measured AHI, it appears that stand-alone use of a Type IV sleep study is not always sufficient for clinical practice [9]. However, a Type IV home sleep apnea monitoring device can be an inexpensive and comfortable step toward wider access to diagnosis and treatment of OSA [9]. This thesis reports on a wearable respiration monitoring device for such a purpose. Potentially, this device can be a great replacement for the state of the art respiration monitoring method in all PSG Types.
1.3 Respiration monitoring

The state of the art method for respiration monitoring in sleep studies is measuring the respiration airflow-induced pressure (stagnation pressure) at a stagnation point in the flow in front of the nostrils, or monitoring the respiration temperature with a thermistor.

The exhaled air is at body temperature (i.e. 37°C) and the inhaled air temperature is at ambient temperature. These two temperatures are independent of the airflow. Therefore, measuring the air in front of a nostril is a useful means to measure respiration rate; however, this is not suitable for classifying hypopnea, since a reduced exhalation flow still is in body temperature and cannot be distinguished from normal exhalation. It has been reported that respiratory airflow measurement via thermistors or thermocouples is inaccurate under conditions typical of sleep studies [21], and misses over 20 percent of hypopnea events [22].

By measuring the stagnation pressure of the respiratory airflow, one can effectively diagnose OSA and distinguish between apnea and hypopnea. It is employed in many commercial HSAT devices. The pressure transducer is placed in a central device mounted on the patient’s chest and is connected to a nasal cannula that goes around the ear, into the nostrils, as shown in Figure 1. The transducer measures the stagnation pressure at the nostrils while the patient breathes. It has been reported that the tube causes discomfort and may dislodge during sleep [23].

In the literature, there are various other methods of respiration monitoring such as humidity measurement [24]–[29], additional thermal-based methods [30]–[33] and drag-based [34]–[38] flow measurement.

Exhaled air has a much higher relative humidity (RH) than inhaled air, which has the same RH as the ambient air. This change in RH can be used to monitor the respiration rate [39], [40]. There are different principles to sense humidity, all incorporate a porous medium that changes some of its mechanical or electrical characteristics due to the absorption of water molecules as the environment becomes more

![Figure 1. Schematic of the state of the art respiration monitoring via nasal cannula](image-url)
humid. A disadvantage of humidity sensors is that they cannot respond fast enough to follow respiration, and even though they can distinguish the presence of breathing from its cessation, their output is not a measure of airflow, which fails to classify apnea and hypopnea.

Thermal-based flowmeters or thermal anemometers measure the velocity of a fluid by sensing the variations in heat loss from a small resistive heater due to convection. There are several types of these sensors [30]–[33] but the presence of a hot element, i.e. the heater, is a serious drawback as it makes them hazardous when placing in the face of a patient in bed. The temperature of the hot element is usually above 70 °C and the sensitivity of the sensor generally increases with that temperature. Additionally, the hot element makes the sensor too power-demanding (usually 12-20 mW) for a face-borne wearable device with limitations on space and weight for the power-supply.

Some very sensitive drag-based flowmeters exist that are inspired by the receptor hairs of insects. They work on the principle of a pillar [41] or cantilever [42] deformation due to the drag force induced by the airflow. These sensors have very high sensitivity and short response time, which makes them suitable candidates for low-range high-frequency flow-sensing, such as hydrophones. However, these types of devices contain fragile moving structures that could easily be damaged by a patient tossing and turning in bed, thus making them unsuitable for PSG.

Some non-contact respiration monitoring devices exist that rely on thermal imaging [43]. With this method, the temperature of the exhaled breath, and the neck temperature are monitored as an indicator of breathing for detecting sleep apnea. The shortcoming of this method is that the sight of the thermal camera is lost if the patient moves, or turns in bed.

Among all these methods a pressure sensor for monitoring respiration for sleep study appears to be the best solution. Further, we will discuss available options for handling the recorded data by the pressure sensor.

1.4 Data communication

Even though, making a full monitoring device with data logging system is beyond the scope of this project, it is instructive to consider the entire system to understand the requirements for the sensing device. Figure 2 shows the methods discussed in this section.
The device will have three main parts: the sensing module, the power-supply unit, and the data logging unit. The sensing module consists of the sensor and the data processing unit. The power-supply unit can be a battery for active setups or an energy harvester such as an electromagnetic coil. Some fully passive setups exist that are explained in Appendix 1. The data logging unit can take different forms but in general, it can be categorized as one of two types: (1) recording the data in the device or (2) transferring it to a distant unit wirelessly.

There are many options for storing the data on the device such as using secure data (SD) cards, flash memories, and random access memory (RAM) ICs. These solutions are relatively bulky and heavy for a wearable device. The data recording unit can be replaced by a data transmission unit. For a portable, comfortable, and easily-wearable sensor, it is crucial that the data transmitter be low power and wireless. The typical communication standards for implantable and wireless sensors are medical implant communications services (MICS), wireless medical telemetry services (WMTS), Bluetooth (BLT) (IEEE 802.15.1), and ZigBee (IEEE 802.15.4) [44]. For these standards, algorithms for optimizing the power consumption of data transmission are available. Active wireless communication can be a suitable solution for a wearable system. In case the power supply unit should be abandoned to further miniaturize the system, there are also passive solutions available.

Some sensor systems are capable of harvesting power from the surrounding ambient and are used in implantable medical devices for drug delivery or monitoring biological parameters [45]. These technologies have a very short range of communication of less than 10 cm and are not suitable for sleep monitoring.
A different technology uses passive long-range communication such as surface acoustic wave (SAW) transmission. SAW technology allows passive and wireless sensors that enable a compact light-weight device design. A more in-depth review of how SAW is used for this purpose is available in Appendix 1.

1.5 Pressure sensors review

Later in chapter 3, it will be explained why ±30 Pa is the desired range of gauge pressure for monitoring respiration. An ideal sensor for sleep apnea detection should be able to distinguish normal respiration, hypopnea, and apnea from each other. Apnea and hypopnea are defined as at least 75% and 30% reduction of respiratory airflow, respectively [46]. According to the Bernoulli equation

\[ P_1 + \rho gh_1 + \rho \frac{V_1^2}{2} = P_2 + \rho gh_2 + \rho \frac{V_2^2}{2} \]

Where \( \rho \) is the density of the fluid, \( h, V, \) and \( P \) are the height, velocity, and pressure of the fluid at the point of interest, respectively, and indexes 1 and 2 indicate two different points, a 75% drop in airflow speed \( V_1 \) yields a ~94% drop in the stagnation pressure \( (P_2 - P_1) \) for \( V_2 = 0 \), \( h_1 \) the height of reference point 1, \( h_2 \) the height of the stagnation point, \( \rho \) the density of air and \( g \) the acceleration of gravity. A pressure sensor with a resolution of less than 6% of the stagnation pressure for normal respiration, i.e. 1.8 Pa can distinguish apnea, hypopnea and normal respiration.

In general, pressure sensors measure absolute or differential pressure. Absolute pressure sensors have a perfect vacuum cavity as a reference, whereas some sensors have a sealed cavity with a known pressure and compare the measured target pressure to the known pressure of that cavity. Differential pressure sensors do not have a cavity with known pressure; they compare the pressure of two sides of their membrane. If one side of the membrane is open to ambient atmospheric pressure, it is said to measure gauge pressure.

Pressure transducers commonly employ some sort of membrane which transduces the input pressure into an electrical signal. Membrane-based pressure sensors are commonly resistive or capacitive. In resistive sensors, the membrane is configured to include a conductive pattern; and when the membrane deforms due to a change in the pressure difference between its two sides, this leads to a change in the resistance of the pattern. The input pressure can then be deduced directly from the change in resistance.
In a membrane-based capacitive pressure sensor, one of the electrodes is a flexible membrane that deflects due to a pressure difference between its both sides, which results in a change in the gap length between the membrane (the moving electrode) and the fixed electrode. In comparison to resistive sensors, capacitive ones are usually more complex to fabricate [47], less sensitive to temperature and have a lower power-consumption [47]–[49]. Also, a capacitive sensor is well suited for a passive impedance-loaded SAW device (see Appendix 1).

Pressure sensors are used in many industrial and medical applications and therefore are designed and fabricated in many ways that suit the particular application the best. Among them, flexible sensors are used in different applications such as electronic skins (E-skin), and wearable sensors due to their good conformal coverage [48], [50], [51]. Flexible pressure sensors usually measure differential or gauge pressure, as they do not have a rigid body for a cavity at reference pressure. They are mostly made of polymers, such as polydimethylsiloxane (PDMS), or polyurethane and most of the fabrication steps do not require a cleanroom environment. In general, the elastomer-based fabrication is less costly compared to the silicon-based fabrication [52], [53]. The capacitance between two parallel electrodes is calculated as

\[ C = \epsilon_0 \epsilon_1 \frac{A}{d} \]  

(2)

where \( \epsilon_0 \approx 8.85 \times 10^{-12} \) F/m is the vacuum permeability, \( \epsilon_1 \) is the dielectric constant of the dielectric material between the electrodes, \( A \) is the overlapping area of both electrodes, and \( d \) is the gap between them, here assumed to be constant across \( A \). We introduce two sensitivities to describe the performance of the sensors. The absolute sensitivity is the ratio of capacitive change \( \Delta C \) to the pressure change \( \Delta P \), and the relative sensitivity is the ratio of the relative change of capacitance \( \Delta C/C_0 \) to the pressure change \( \Delta P \), where \( C_0 \) is the capacitance at zero pressure.

In most cases, the dielectric material between the capacitor plates is air or vacuum, and the membrane is suspended along its edge. In Appendix 2, we discussed the physics of suspended membranes, but to mention shortly here, the deflection of a suspended membrane in a specific pressure difference between its both sides, increases with lower stiffness of its material, lower thickness, and larger area of the membrane. Furthermore, the maximum deflection of the membrane must be smaller than the gap length of the capacitor. By using stiff materials such as silicon, or SU-8, researcher could make capacitors with suspended membranes with high ratio of area to gap length, which results in a high capacitance. Capacitive sensors are prone to random noises coupled into the system through external features such as connecting wires. By having a sensor with high capacitance, a high signal to noise ratio
(SNR) is followed [47]. Whereas, by using relatively less stiff materials such as PDMS, due to fabrication feasibility limitations, a much lower area to gap length ratio can be reached for a fully suspended membrane, that yields a capacitor with either wide gap or small area. In both cases, the capacitance would be low, and hence the SNR. To overcome this problem, some researchers have employed a solid material with a higher dielectric constant as the dielectric layer to fill the cavity and support the membrane.

Cui et al. [48] used a “one-dimensional pyramid patterned PDMS structure” for higher flexibility of the electrodes and filled the gap with polystyrene microspheres as dielectric, as shown in Figure 3. The sensor demonstrates a relative sensitivity of 0.741 kPa⁻¹ for the range of 0 to 1 kPa and could tolerate bending by up to 120° degrees without breakage. Ma et al. [51] implemented a one-dimensional pyramid PDMS structure as the dielectric between two flat electrodes. This design showed a relative sensitivity of 2.04 kPa⁻¹ up to 2 kPa pressure with a resolution of 7 Pa. These two sensors have an area of ~1 cm², and are 3.5 and 4.5 pF at zero (gauge) pressure, which are small capacitances that yield a low SNR. Using an elastic dielectric may seem beneficial for its higher dielectric constant but it limits the deflection of the membrane.

![Figure 3. Schematic of the Cui et al. design [48]](image)

Other researchers have used an assembly of parallel capacitive sensing modules [47], [49]. A single capacitive pressure sensor module shows good sensitivity but has small capacitance, hence low SNR. One can increase the capacitance by connecting those single capacitive sensors in parallel. The parasitic capacitance can be limited by having the modules share the same electrodes, instead of connecting them externally. Such an assembly will have the same relative sensitivity with a higher absolute capacitance, hence a higher absolute sensitivity. Another benefit of this design concept is that the design is extendable to the desired capacitance.
In 2009, Pederson et al. [47] designed a capacitive pressure sensor with such parallel capacitors. A support structure with hexagonal pattern was used to hold the silicon membrane suspended. The gap was 420 nm and diagonal of the hexagons was 150 μm. The diameter of the entire circular membrane was 2.3 mm. With this geometry the total sensor capacitance was 106 pF at atmospheric pressure and the membrane touched the bottom electrode at 200 kPa. A schematic of a quarter of the sensor is depicted in Figure 4.

![Schematic of sensor](image)

*Figure 4. A quarter of the sensor presented by Pedersen et al. [47]*

In 2017, Berger et al. [49] designed a capacitive pressure sensor with the same concept of parallel capacitors. The graphene membrane was laid over a 3D cavity structure. The structure was a layer of SiO$_2$ with an array of hexagonal holes with diagonals of 30 μm and depths of 180 nm fabricated by deep reactive ion etching (DRIE) process. Therefore, only the parts suspended over the holes can deflect due to a pressure difference. The full pressure scale for this sensor is 0 to 100 kPa with an absolute sensitivity of 4.2 pF/kPa.

Both of these works were designed for higher pressure ranges than required for respiration monitoring during sleep, and their fabrication process used costly silicon microfabrication methods such as DRIE.

1.6 Objectives

There is a need for a more comfortable device for respiration monitoring to be used as a stand-alone Type IV HSAT, or a replacement for current respiration monitoring method in other Types of PSG. As shown in Figure 5, this device will be mounted above the upper lip, with short tubes that are placed at
the nostrils along the airflow direction, to sense the stagnation pressure for respiration monitoring. In this work, we propose a sensor design for that purpose. The stagnation pressure is measured to cover the range of ±30 Pa (gauge pressure) in nasal tidal breathing with a resolution of at least 3.5 Pa to be able to clearly classify apnea and hypopnea events. However, the sensor should be robust to overpressure and under-pressure scenarios such as deep breath, or sternutation. Therefore, a range of -50 to 100 Pa would give a good safety margin for measuring normal and reduced respiration of different people. The sensor must be able to follow the respiration rate from 12 to 28 breaths per minute (0.2 to 0.47 Hz) [54]. As mentioned, the sensor is supposed to be mounted on the upper lip, so it needs to be small and flexible to provide the most comfort for the patient. It also should be flat on the surface of the skin in a way that does not block the nasal airflow so the patient will not feel an external object in front of the nostrils. Based on the anatomy of the face, the sensor size should not exceed 1 cm² for conformal coverage and patient’s comfort.

![Figure 5. A schematic of the respiration monitoring device on the patient’s upper lip](image)

We compared different methods of respiration monitoring in section 1.3, and decided to continue with the state of the art pressure measurement method. We will design a capacitive sensor, for its low sensitivity to temperature, and low power consumption compared to resistive pressure sensors. A modular design will be suggested for the membrane-based capacitive sensor that results high sensitivity and high SNR.

The fabrication processes and materials should be chosen for a relatively low-price sensor. In this work, only the development of the wearable pressure sensor is explored. However, it is nonetheless important to consider the applicability of our design with respect to the transmitter design, which can be implemented at a later stage.
CHAPTER 2: MICROFABRICATION

The sensor is made of two main parts that are fabricated separately and then bonded together. In Figure 6, the whole fabrication process is depicted in summary. The fabrication process consists of two parts, fabricating the membrane and the base. The base consists of three layers of PDMS and a layer of Parylene-C. An SU-8 2005 (Kayaku Advanced Materials Inc.) mold is fabricated on a glass substrate using photolithography techniques. Then a thin layer of diluted PDMS (with hexane) is spin-coated over the mold to form the insulation layer and the spacers that define the capacitor gap. After that, a conductive PDMS electrode, infused with carbon black (CB), is bonded to the lower layer. At last, a layer of PDMS is cast over the sample. To fabricate the membrane, a layer of PDMS is spin-coated over a sacrificial layer of photoresist. Then gold electrodes are deposited onto the PDMS layer through a shadow mask. For the assembly of both parts, the base is peeled off from its mold and bonded to the membrane. Finally, the sacrificial layer is developed to release the sensor.
2.1 Fabrication of the membrane

The membrane is a 10 μm PDMS layer with gold electrodes that are deposited by sputtering. To fabricate this membrane, a glass slide is coated by AZ5214 photoresist at 1000 rpm for 30 seconds and then soft-baked on a hotplate. The thickness of this layer is not crucial for the rest of the fabrication process. AZ5214 is a positive photoresist and in this process, it is used as a sacrificial layer that will be fully developed later to release the sensor from the substrate. Therefore, it should be UV-exposed (UV-KUB 2, Kloe Inc.) for 4 seconds at an intensity of 50 mW/cm² after spin coating. The PDMS (Sylgard 184, Dow) is
prepared by mixing base and hardener at a mass ratio 10:1. A 10 μm PDMS layer is spin-coated on top of the sacrificial layer in two steps. First at a speed of 500 rpm for 10 seconds after an acceleration of 100 rpm/s, and then at a speed of 2500 rpm for 5 minutes after an acceleration of 200 rpm/s. The sample is cured on a 90°C hotplate for 1 hour. Figure 7 shows the spin coating curve for PDMS with 10:1 ratio of resin to curing agent.

![Spin coating curve of PDMS (10:1) for 5 minutes in different speeds](image)

*Figure 7. Spin coating curve of PDMS (10:1) for 5 minutes in different speeds*

It should be noted that the linear coefficient of thermal expansion of PDMS (3.2×10⁻⁴ K⁻¹ [55]) is significantly higher than that of glass (9×10⁻⁶ K⁻¹ [56]). As a result, curing it on these layers at elevated temperature yields a tensile residual stress in the PDMS after cooling down to room temperature. This residual stress can help keep tension in the membrane when it is bonded to the base and released from the substrate.

A shadow mask is made from aluminum foil for selective metal deposition of the electrodes on the membrane. The mask is prepared by cutting pieces out of the aluminum foil using an executor knife. The metal deposition occurs via sputtering, which limits the increase in the sample’s temperature compared to other deposition methods such as e-beam evaporation. The shadow mask is placed on the sample before transferring it into the sputter coater (Nexdep Sputtering, Angstrom engineering Inc.). Deposition of a 5 nm titanium adhesion layer is followed by 50 nm of gold. Titanium oxidizes very easily; for this
reason, both metal layers are deposited in one session using two separate targets in the sputter coater without breaking the vacuum between the deposition steps. More details about the sputtering settings are provided in Table 1.

Table 1. Sputtering parameters

<table>
<thead>
<tr>
<th>Material</th>
<th>Final thickness (nm)</th>
<th>Deposition rate (nm/s)</th>
<th>Power source</th>
</tr>
</thead>
<tbody>
<tr>
<td>Titanium (Ti)</td>
<td>5</td>
<td>0.05</td>
<td>RF</td>
</tr>
<tr>
<td>Gold (Au)</td>
<td>50</td>
<td>0.2</td>
<td>DC</td>
</tr>
</tbody>
</table>

The thickness of the deposited layers is monitored by a built-in sensor inside the machine. To verify the layer thickness through a separate measurement, a dummy glass slide, partially masked by a narrow piece of tape, is sputter coated alongside the main samples. After the deposition, the tape is removed from the dummy sample and the actual accumulated thickness of the metal layers is measured using interferometric profilometer (Profilm 3D, Filmetrics Inc.).

Even though sputtering does not heat the sample much, it appears that the PDMS surface must have experienced an increased temperature during the process; due to the higher coefficient of thermal expansion of PDMS (3.2×10⁻⁴ K⁻¹ [55]) compared to gold (1.4×10⁻⁵ K⁻¹ [57]), its surface appears to expand while the metal is deposited. After the deposition, it cools down and shrinks more than the metal coating,
resulting in random patterns on the surface of the sample with a characteristic length scale. Figure 9 shows the top view of the metal deposited membrane taken by optical microscopy.

![Figure 9: Patterns on the surface of deposited metal layer on PDMS. a. 500 nm of Aluminum deposited on PDMS layer by e-beam evaporation. b and c. 5 nm of titanium and 50 nm of gold deposited on PDMS layer by sputtering.](image)

The size of the wrinkles is approximately 2.6 ±0.4 μm in wavelength and 0.25 μm in amplitude (peak to peak). The values are averaged from 14 measurements using interferometry surface profiler. Figure 10 shows the top view of the post-processed result of the wrinkles’ profile. Green and yellow areas are deeper and higher parts of the surface, respectively. Each measurement is taken at the shortest distance between two parallel green lines.

![Figure 10: Gold and titanium on PDMS wrinkle size measurement using Profilm3D surface profiler](image)

### 2.2 Fabrication of the base

The base contains the CB-PDMS electrode and the spacers and is fabricated layer-by-layer on a glass substrate. We used 4 inch glass wafers (Economical glass slide, Ted Pella Inc.) for the substrate.
2.2.1 Mold fabrication:

In the cleanroom, the substrate is rinsed with acetone, followed by IPA. Then it is dried with the aid of a nitrogen gun in the fume hood, and it is placed inside a 160°C oven to dehydrate for 15 minutes. Next, it is left in the fume hood to cool down to room temperature. A thin layer of OmniCoat (Kayaku Advanced Materials Inc.) is spin coated on the surface of the glass at a speed of 3000 rpm for 30 seconds with acceleration of 300 rpm/s, to improve the adhesion between glass and SU-8. After spin coating, the substrate is placed on a hotplate at 200°C for 5 minutes. After taking it off the hotplate, it cools down to room temperature.

Two other methods for improving the adhesion of glass and SU-8 were shown to be unsuccessful. The first method involved spin coating a layer of hexamethyldisilazane (HMDS), a common adhesion promoter, onto the substrate but the results were not satisfying as in subsequent fabrication steps the SU-8 was peeling off from the substrate with the PDMS layers. The second method uses a double-layer of SU-8. In this method, the main patterned SU-8 layer is deposited on top of an unpatterned layer of SU-8. This method uses double the amount of SU-8, it is time-consuming, and very delicate; the success rate of this method was not more than 70%.

The next step is to spin coat a 5 μm SU-8 2005 layer on the wafer using a three-step recipe. First, the SU-8 is spread on the whole surface by spinning at speed of 500 rpm for 10 seconds after a 100 rpm/s acceleration. This is followed by spinning at 4000 rpm for 30 seconds after a 300 rpm/s acceleration to reach the desired thickness. Finally, the speed is returned to zero with a deceleration rate of 1000 rpm/s.

When the spin coating is done, the substrate undergoes a two soft baking steps, 1 minute at 65°C, followed by 3 minutes on a 95°C hotplate. It is recommended to transfer the substrate from the spin coater to the hotplate by holding it from two opposite edges with fingers. This prevents surface unevenness in the SU-8 layer that would be caused by wafer tweezers, as any unevenness on the surface of the SU-8 will cause an unwanted gap between the mask and the sample in the photolithography step. The other benefit is that due to the geometry of the spin coater, it is impossible to take the substrate off the spinner head using wafer tweezers without tilting which can be another source of unevenness in the thickness of the SU-8 layer, as SU-8 2005 has a very low viscosity. The two-step baking helps reduce the local stresses caused by fast temperature ramping, hence resulting a better adhesion between the SU-8 mold and the substrate.
After leaving the sample to cool down, it is ready for UV exposure (NXQ 4006, Neutronix Quintel). The substrate is placed on the mask aligner’s chuck with the SU-8-coated side facing up, and the chrome on glass photomask is loaded into the mask aligner. There are 7 samples on each wafer.

The photomask is aligned with the edges of the wafer, with the printed side touching the sample. The intensity of UV exposure in the mask aligner device is 19.4 mW/cm². Initially, a transparency mask was used for the fast turn-around time between ordering and receiving it, but this approach was discontinued for the benefit of glass-based chrome masks due to the micro-bubbles in the transparency sheet that affect the final surface of the SU-8. Figure 11 is a comparison of two identical SU-8 samples, one exposed without any photomask (flood exposure) and the other with an unpatterned (clear) transparency mask. The measurement is taken by an optical profilometer (WYKO NT1100, Veeco Inc.), and the figure is the 3d model generated in its software. The figures are the top view of the measured surface and the lighting is set to illuminate from top right of the surface in a way to show the roughness of the surface clearly, therefore colors are not representative of depth. The deepest bumps were measured to be less than 0.5 μm.

Figure 11. Effect of micro-bubbles in the transparency mask on SU-8 surface. a. Without photomask. b. With a blank photomask figure taken by Wyko interferometry surface profiler

After the exposure, the sample undergoes a post exposure baking (PEB) process (1 min on 65℃ and 3 min on 95℃). By the end of the PEB, the patterns will be visible in the SU-8 layer. When the slide is cooled down, it is soaked in a bath of SU-8 developer for 10 seconds and then rinsed very shortly with IPA followed by a generous rinse of deionized water. After drying the slide with a nitrogen gun, the process is repeated, this time for only 5 seconds. White stains after the first developing step show there is still
underdeveloped SU-8 on the sample, but a two-step developing process has been shown experimentally to bear a lower risk of over-developing the pattern. Table 2 shows a summary of the fabrication recipe for fabricating the SU-8 mold.

It should be noted that the OmniCoat layer will not be removed from the glass, but it stays between the glass and the cured SU-8. The thickness of this layer is not important and has no significant effect on the rest of the fabrication process. Overdeveloping the OmniCoat layer during the SU-8 developing step can cause undercut for the SU-8 layer.

Table 2. SU-8 mold fabrication recipe

<table>
<thead>
<tr>
<th>Photoresist</th>
<th>Recipe 1</th>
<th>Recipe 2</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>SU-8 2005</td>
<td>SU-8 2005</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Photomask</th>
<th>transparency</th>
<th>glass</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thickness required</td>
<td>5.5 ± 0.5 μm</td>
<td>5.5 ± 0.5 μm</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Spin coating (speed/acceleration/duration)</th>
<th>Recipe 1</th>
<th>Recipe 2</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>500/100/10 +</td>
<td>500/100/10 +</td>
</tr>
<tr>
<td></td>
<td>4000/300/30 +</td>
<td>4000/300/30 +</td>
</tr>
<tr>
<td></td>
<td>0/500/0</td>
<td>0/500/0</td>
</tr>
</tbody>
</table>

| Soft bake                     | 1 min on 65°C +   | 1 min on 65°C + |
|                              | 3 min on 95°C     | 3 min on 95°C   |

| UV intensity                 | 19.4 mW/cm²       | 19.4 mW/cm²     |

| UV exposure                  | 7.5 seconds       | 3 seconds       |

| Post exposure bake           | 1 min on 65°C +   | 1 min on 65°C + |
|                              | 3 min on 95°C     | 3 min on 95°C   |

| Develop                      | 10 s + IPA/water rinse | 10 s + IPA/water rinse + 5 s + IPA/water rinse |

2.2.2 Base layers

Before spin-coating/casting PDMS on the mold, an adhesion reduction treatment using silanization HMDS is performed. Without HMDS treatment of the SU-8 surface, adhesion between PDMS and SU-8 can be that high that either PDMS spacers get stuck in the SU-8 holes and break while peeling off the base, or the SU-8 peels off from the glass substrate while peeling off the PDMS sample. The surface treatment with HMDS can be done in vapor or liquid phase. For the former, the slide is put in a petri dish and a few
droplets of HMDS are deposited around the slide inside the petri dish. The petri dish is put in a vacuum chamber and kept at low pressure for 5-7 hours. This method works perfectly, as long as the chamber seal can hold the vacuum. However, HMDS vapor damages the chamber’s O-ring and after a few months of using the chamber for HMDS treatment, the chamber is unable to keep the vacuum. For that reason, the liquid phase method is used instead. In this method, the whole surface of the mold is covered with HMDS, and it is left in the fume hood to evaporate at room temperature. This volatile chemical has a boiling temperature of 65°C and evaporates in less than an hour. The liquid phase method is easier but uses more HMDS that is a toxic material and should not be inhaled.

The first layer is a 1 μm PDMS layer that has the spacers and acts as an insulation layer that prevents short circuit between electrodes in case of over-pressure. As we discussed in section 3.3.1, this layer should be as thin as possible. The lowest thickness we could reach for PDMS was 8 μm. Thus, for this layer, diluted PDMS 7:1 ratio (Hexane:PDMS) is used. To make such mixture, 0.3 g of PDMS (10:1 ratio base to hardener) is added to an empty glass container, and then 2.1 g of hexane is added. The initial mixing is done by shaking the container after closing its lid. Then it is loaded into a planetary mixer (ARE-310, Thinky Inc.) to mix it for 3 minutes, followed by 2 minutes of defoaming. After that, the diluted PDMS is poured on the previously made SU-8 mold and is spin coated (Laurell WS-650-23) at 5000 rpm for 2 minutes after an acceleration of 500 rpm/s to form a 1 μm layer. At lower acceleration, the PDMS layer is thicker as the hexane evaporates before reaching the final speed. Figure 12 shows the spin coating curve for diluted PDMS at different Hexane-to-PDMS ratios.

![Figure 12. Spin coating curve of diluted PDMS with different ratios at 5000 rpm for 120 seconds after 500 rpm/s](image)
The diluted PDMS is thin enough that there is no need for degassing it to make sure it has filled the mold. Degassing causes hexane to evaporate which changes the ratio and hence the final thickness.

After spin coating a thin layer of diluted PDMS on the mold, since the final thickness of the PDMS layer is smaller than the depth of the mold, the holes are not completely filled. During the curing time, PDMS from surrounding of each hole flows into it but due to high viscosity, and limited mass of PDMS, a very thin layer stays at the edge of the hole. This will cause the spacer to get stuck in the mold while peeling off the sample, as shown in Figure 13.

We designed and tested three different spacer shapes to study this effect; cylindrical spacer with a diameter 20 μm, cylindrical spacer with a diameter 50 μm, and cylindrical hollow spacer with an inner and an outer diameter of 20 and 50 μm, respectively. Figure 14 shows the profile of the PDMS surface measured by the optical profilometer, and it shows that the holes are just partially filled with diluted PDMS. The figures are drawn by overlaying two surface profiles taken separately using the Wyko optical surface profiler. They were taken from the surface of the SU-8 mold before spin coating the diluted PDMS and from the surface of the spin coated diluted PDMS; the dotted lines are added arbitrarily for better visualization. This thinning effect was the least and almost negligible in the 20 μm diameter mold design; Therefore, we continued with that.
a. 20 μm mold

![Graph showing the height profile for a mold with a diameter of 20 μm.]

b. 50 μm mold

![Graph showing the height profile for a mold with a diameter of 50 μm.]

c. Hollow 50 μm mold

![Graph showing the height profile for a mold with an inner diameter of 20 μm and an outer diameter of 50 μm.]

*Figure 14. a. Mold with diameter of 20 μm b. Mold with diameter of 50 μm. c. Mold with inner diameter 20 μm and outer diameter of 50 μm.*
The electrode in the base is designed to be sandwiched between the 1-µm-PDMS layer and a 2 mm (or thicker) layer of PDMS. The conductive PDMS is made by mixing PDMS and carbon black (CB) powder with a certain ratio. Too little amount of CB is not enough for having a conductive mixture and too much CB prevents the mixture from curing. We used a ratio of 5:1 (PDMS:CB) by adding 1 g of CB and 5 g of PDMS to a clean empty glass container and mixing in the mixer for 3 minutes at 2000 rpm followed by 2 minutes at 2200 rpm. To measure the conductivity of the mixture, we prepared 5 conductive PDMS samples with dimensions of 3 by 1 by 0.1 cm with two wires with spacing shown in Figure 15. The electrical resistance between the wires is measured by a multimeter (VC830L, Victor Inc.) to be 2.05±0.05 Ω. The fact that the resistance of 5 geometrically-similar samples of conductive PDMS is identical, proves that the distribution of CB in the conductive PDMS mixture was uniform at the mm-scale. The resistivity of the samples was estimated at $\rho_{CB} = 10^{-3} \text{ Ω m}$.

![Figure 15. Schematic of the conductive PDMS samples for uniform distribution test (dimensions in mm)](image)

To make the CB electrodes, the mixture is spread with thickness of 200 µm on clean glass slides using another glass slide and spacers with known thickness. After curing on a 60°C hotplate for 4 hours, stripes of 1 cm width are cut from this sheet, each further cut to make 4 electrodes as depicted in Figure 16. A hole is punched in the middle of each electrode using a hole punch with outer diameter of 1.5 mm to eliminate the risk of short circuit from the air escape hole, as explained later. Conductive PDMS is also used to make 2 by 2 by 5 mm parts (here called CB rod) for electrical connection to the CB electrode. CB rods are preferred to metal wires, because they can bond to the CB electrode and also the thick PDMS base layer using plasma bonding.
Figure 16: CB sheet to be cut into electrodes (left) and electrode with hole punched in its center (right).

The sample, which has the diluted PDMS layer, and CB electrodes are put in the plasma chamber and treated for 60 seconds at high power (30 W) with house compressed air. CB electrodes are put in the chamber with the smoother surface facing up. Under the microscope, each electrode is carefully aligned with the patterned SU-8 using fine tweezers. Bubbles trapped between the electrode and the 1-µm-PDMS layer are gently pushed to the sides by a cotton-tipped applicator. Then after another plasma treatment, a CB rod is bonded to each electrode. As this face of the electrode is rough, there’s a high chance of poor bonding between the rod and the electrode. The resistance between the electrode and the end of the rod is measured by a multimeter. As the electrical resistance is too high (+200 kΩ), conductive silver paint (Circuit Scribe) is added to the root of the rod, as shown in Figure 17.

Figure 17. A complete sample with reinforced connection using silver paint
In this step the electrode is sandwiched with a thick layer of PDMS. After activating the surface of the sample, PDMS is cast over the sample and degassed so there will be no bubbles trapped in it. The thickness of this layer is not crucial, but it should not pass the height of CB rods. The sample is cured on a 60°C hotplate for 4 hours. When the sample is cooled down to room temperature, the sample is peeled off from the mold, then a through hole with diameter of 2.5 mm is punched with the side with spacers facing up, for access to the gold electrode after bonding the lower part to the membrane (see Figure 6.A5). Another smaller hole is punched in the middle of the hole that we made in the CB electrode, to allow air escape from between the capacitor plates. The position of both the holes are shown in Figure 18.

![Figure 18. Left) Spacers with 600 μm spacing on the surface and the hole in the CB electrode with diameter 1 mm. Right) Schematic of the holes punched in the peeled base part before bonding to the membrane](image)

The last layer that is added on top of the base part before bonding it to the membrane is the Parylene-C coat. The surface is covered with Kapton tape with the electrode surface open. The samples are then put in the Parylene coater to coat a 0.9 μm thick layer of Parylene-C on the spacers and the thin PDMS layer on top of the CB electrode. This layer prevents membrane collapse, i.e. the adhesion of the gold electrode on the membrane to the base part, as visually observed.

2.3 Bonding the base to the membrane

To bond the base to the membrane, both parts are placed in the plasma chamber (PDC-001, Harrick plasma), with the bond surfaces facing up. They are treated for 1 minute with 30 W power and immediately bonded together while aligning the two electrodes under the microscope. A piece of wire is attached to the gold electrode using silver ink through the hole that was punched into the base. Then the hole is filled with PDMS and left to dry. Using plastic tweezers, we scratch the membrane between the
samples on the substrate to facilitate the development of the sacrificial layer, as shown in Figure 19. AZ MIF-300 developer is poured slowly on the scratches using a pipette. It takes 20 minute for the sample to be released but if the sample does not slide freely on the substrate, it may need some more time for development. Adding fresh developer helps, as well. Developer should not reach the air escape hole. The contact angle between developer and PDMS is high making it easy to control the droplets.

![Image of membrane with scratches](image)

*Figure 19. Scratches on the membrane layer for easing the development step*

The membrane is supposed to be suspended over 313 spacers. The initial deflection of the membrane between two spacers with diameter of 20 μm with spacing of 240 μm can be seen in Figure 20. It was taken by an optical profiler (Profilm3D, Filmetrics) and shows the profile of the outer side of the membrane suspended between two spacers. In the ideal case, with no initial deflection at atmospheric pressure, the gap would be 5 μm everywhere under the membrane.
For a 600 μm spacer pitch, we could not get any measurement from the suspended membrane surface profile, since it was not stretched and flat enough for optical profilometry. In fact, the membrane was loose right after the sacrificial layer was developed and the sample was released, as shown in Figure 21. There are three possible reasons for the looseness of the membrane. One is that it is mechanically stretched in the process of developing the sacrificial layer and removing the sample from it. Second, the developer swells the PDMS membrane and loosens it. Three, the residual compression stress in the wrinkled metal electrode is released when the membrane is released from the substrate and loosens the membrane. Not enough study and observation were done on this problem, however, the third reason sounds more probable.
Figure 21. The loose membrane right after being released from the substrate by developing the sacrificial layer

The photos of the final fabricated sample is depicted in Figure 22.

a.  

b.  

Figure 22. Both sides of a final product
In this chapter, we discuss the different aspects of the sensor design, and optimize the shape and dimensions of the sensor features.

3.1 Range of pressure

To design the sensor, one of the main factors that should be set is the target range of pressure. The stagnation pressure of the nasal airflow of a healthy person (i.e. the author, Hamed Pouriayevali) during night sleep is measured via a Type III HSAT device (ApneaLink Air, ResMed) and is shown in Figure 23.

Human respiration has several modes. Tidal breathing is when one breathes without extra effort. The tidal volume is the volume of air displaced between a normal inspiration and a normal expiration and it measures about 500 ml or 7 ml/kg of body mass [58]. The respiration rate has been reported to be between 12 to 28 breaths per minute for adults and elderly people [54]. Therefore, the period of a full respiration cycle is set to be 3 seconds. To calculate the airflow-induced pressure range, a sinusoidal model is assumed for breathing with a period of 3 seconds as shown in Figure 24.
The tidal volume is the area under the sinusoidal volumetric flow curve, and is calculated as

$$\text{tidal volume} = \int_{0}^{T/2} Q_m \sin \left(\frac{2\pi t}{T}\right) dt$$

(3)

in half a cycle where $T = 3\, s$ is the period and

$$Q_m = A_{\text{nostrils}} V_m$$

(4)

is the maximum volumetric flow in $\text{m}^3/\text{s}$. Assuming $A_{\text{nostrils}}$ being the area of the nostrils is $0.75\, \text{cm}^2$, yields the maximum velocity $V_m = 6.98\, \text{m/s}$. The Bernoulli equation (Equation 1) allows measuring the stagnation pressure of respiration while the velocity is maximum. Assuming zero air velocity at the stagnation point ($V_2 = 0$), negligible elevation difference ($\Delta h = h_2 - h_1 = 0$), and the density of air being $\rho = 1.2\, \text{kg/m}^3$, the pressure difference, i.e. the gauge pressure at the stagnation point, will be equal to $29.2\, \text{Pa}$. The analytical solution for the stagnation pressure roughly confirms the gauge pressure range of $\pm 30\, \text{Pa}$ for the design of the pressure sensor for monitoring respiration. For the extreme cases of 12 and 28 bpm, the pressure is calculated as $17.5\, \text{Pa}$ and $40.9\, \text{Pa}$, respectively, which may affect the requirements of the sensor. However, for all the requirements, we considered the average case of $30\, \text{Pa}$ pressure.
3.2 Sensor design

The objective is to make a capacitive pressure sensor to measure respiration airflow-induced pressure that changes in a range of ±30 Pa gauge pressure. In section 1.5, we discussed the modular design of a capacitive pressure sensor. The 1×1 cm² membrane is suspended over a pattern of cylindrical spacers. The pattern is designed in a way to partition the membrane into sections with similar support (boundary condition) underneath.

We compare pattern A with spacers at the vortices of regular hexagons with pattern B with spacers in the middle of regular hexagons, both shown in Figure 25. The height and spacing of the spacers are the design parameters and are not independent as the spacing regulates the maximum deflection and the height regulates the initial gap (height of cavity under the membrane). Independent of the spacing between the spacers, the pattern dictates how the module’s membrane deflects. Since the membrane deflection is not uniform, the capacitance of each module is integrated as

\[ C = \int_A \varepsilon_0 \varepsilon_1 \frac{1}{d} \, dA \quad \text{(5)} \]

Figure 26 shows the two patterns with the same module area. Even though both are showing almost the same maximum deflection, for pattern B a much higher portion of the membrane is highly deflected, which results in a greater capacitive change and sensitivity according to Equation 5.
As discussed in chapter 2, the spacers are not bonded to the membrane. Therefore, in negative pressure, where the target pressure is lower than the cavity pressure, the whole membrane is free to be deflected outward of the cavity and be lifted from the spacers. In that case, the deflection problem would be the deflection of a 1 by 1 cm membrane, whereas in positive pressure, due to presence of the spacers under the membrane, it can be seen as a cluster of hexagonal membranes (modules) as is shown in Figure 26.

3.3 Simulations

In this section, the optimum height and spacing of spacers for maximum sensitivity will be derived by simulating the performance of the device. The simulation of the device is done in COMSOL multiphysics v5.5. As explained in chapter 2, the pressure sensor consists of layers of PDMS, Parylene-C, gold, and titanium. Since the geometry of the pressure sensor is a cluster of hundreds of identical sensing modules, to reduce the computational cost, the performance of only one is studied and used to estimate the behavior of the whole pressure sensor. Each sensing module is a regular hexagon with a cylindrical spacer in the middle of it. According to Figure 26.b, due to symmetry, only one sixth of its geometry, i.e. an equilateral triangle, is modeled. The boundary conditions and the uniform loading are shown in Figure 27.b-f. Only the parts that affect the results mechanically or electrostatically are modeled, hence, the lower electrode, i.e. the conductive PDMS electrode and the base layer are not modeled. The layers, their material, and thickness are shown in Table 3. Material and thickness of the layers of the sensor
layer 1 is the top-most and layer 7 is the bottom-most layer. Table 3, where the layers are listed according to Figure 27 with the first row showing the first (top-most) layer of the geometry.

Figure 27. a. Model geometry in COMSOL. b. Symmetry boundary conditions. c. Loading surface. d. Terminal boundary condition with voltage 1v. e. Ground boundary condition. f. Fixed surface boundary condition.
Table 3. Material and thickness of the layers of the sensor
layer 1 is the top-most and layer 7 is the bottom-most layer

<table>
<thead>
<tr>
<th>Layer #</th>
<th>Material</th>
<th>Thickness (μm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>PDMS</td>
<td>10</td>
</tr>
<tr>
<td>2</td>
<td>Titanium</td>
<td>0.005</td>
</tr>
<tr>
<td>3</td>
<td>Gold</td>
<td>0.050</td>
</tr>
<tr>
<td>4 &amp; 5</td>
<td>PDMS</td>
<td>Air</td>
</tr>
<tr>
<td>6</td>
<td>Parylene-C</td>
<td>0.9</td>
</tr>
<tr>
<td>7</td>
<td>PDMS</td>
<td>1.2</td>
</tr>
</tbody>
</table>

Figure 28 demonstrates that the assumption of identical sensing modules is not realistic and in reality, the modules at the edge of the main membrane have different geometries and performance. We also expect small variations in the dimensions of the fabricated sample from the designed geometry, due to fabrication imperfections.

Figure 28. The edge modules (shown in pink) have different boundary conditions than the central modules.

Each module acts as a pressure sensor and as they are all connected to each other in parallel, their capacitance magnitudes adds up. The total area of the electrode is constant (1 cm²), therefore the total capacitance of the sensor is not related to the number of sensing elements or the size of each one.

To set the dimensions of the sensor we need to optimize the gap width and the spacing (pitch) of the spacers. Figure 29 shows the maximum deflection of the membrane with layers given in Table 3 for different spacer spacing at a pressure of 30 Pa, and Figure 30 shows the relative capacitance change.
relative capacitance change = \frac{\text{capacitance at 30 Pa} - \text{capacitance at no pressure}}{\text{capacitance at no pressure}}

of a single sensing module at pressure of 30 Pa compared to initial capacitance at no pressure.

![Figure 29. Maximum deflection at 30 Pa pressure as a function of spacer spacing](image)

![Figure 30. Relative capacitance change of a sensing module for 30 Pa pressure](image)

The range of dimensions is chosen to be feasible to microfabricate and yield a sensor with high sensitivity to pressure changes. In chapter 2, the fabrication process and its challenges were thoroughly
discussed. The final dimensions are selected to be gap of 5 μm and spacer pitch of 600 μm, however, it is recognized that such aspect ratios are challenging to achieve while preventing collapse of the membrane.

3.3.1 Capacitance results verification

Each sensing module consists of several layers with different dielectric constants. So, we can divide one sensing module to 4 sub-capacitors connected to each other as depicted in Figure 31. C1 is the capacitance of one spacer from the lower to the upper electrode that includes the thin layer of PDMS, the PDMS spacer, and the Parylene-C layer on the top surface of the spacer (the Parylene-C layer on the lateral side of the spacers in neglected). It is a cylinder with radius \( r_{\text{spacer}} \) 10 μm and height of 7.1 μm (PDMS: \( d_{sp}=5 \) μm, and \( d_{pd}=1.2 \) μm, where \( d_{pd} \) is the thickness of the thin PDMS layer between spacer and the PDMS electrode, and Parylene-C: \( d_{pr}=0.9 \) μm). \( C_2, C_3, \) and \( C_4 \) are the capacitances of the air-, Parylene-C, and PDMS-filled parts of the cavity, respectively, with geometry of a hexagon with side length of 300/\( \tan 60^\circ \) μm with a cylindrical through hole in the middle (the spacer) and heights of \( d_g \) (ideally equal to the height of the PDMS spacer), \( d_{pr}=0.9 \) μm, and \( d_{pd}=1.2 \) μm, respectively. The relative dielectric constants for air, Parylene-C, and PDMS are 1, 4.4 [61], and 2.75 (based on COMSOL material library), respectively.

\[ C_{\text{module}} = C_1 + C_2 + C_3 + C_4 \]

\[ C_1 = \frac{2\pi \varepsilon_0 \varepsilon_{\text{PDMS}} \ln \left( \frac{r_{\text{spacer}}}{d_{sp}} \right)}{d_{sp} + d_{pd}} \]

\[ C_2 = \frac{2\pi \varepsilon_0 \varepsilon_{\text{Parylene-C}} \ln \left( \frac{r_{\text{spacer}}}{d_{pr}} \right)}{d_{pr}} \]

\[ C_3 = \frac{2\pi \varepsilon_0 \varepsilon_{\text{PDMS}} \ln \left( \frac{r_{\text{spacer}}}{d_{pd}} \right)}{d_{pd}} \]

\[ C_4 = \frac{2\pi \varepsilon_0 \varepsilon_{\text{air}} \ln \left( \frac{r_{\text{spacer}}}{d_{g}} \right)}{d_{g}} \]

Figure 31. a. Cavity of a single sensing module. b. Exploded view of sub-capacitors in a single sensing module. c. Connection of sub-capacitors in each sensing module.
As mentioned earlier, the whole 1 cm² area of the membrane is not covered with hexagonal modules, so, to calculate the total capacitance of the sensor, parameters $C_{2t}$, $C_{3t}$, and $C_{4t}$ are defined that are square prisms with side length of 1 cm and heights equal to $d_g$, $d_{pr}$, $d_{pd}$, respectively, with 313 cylindrical through holes with radius $r_{\text{spacer}}$ for the 313 spacers.

\[
A_{\text{module}} = 6 \times \frac{300^2}{2 \tan 60^\circ} \mu\text{m}
\]  

(6)

\[
A_t = 1 \text{ cm}^2
\]

\[
A_s = \pi r_{\text{spacer}}^2
\]  

(7)

The initial capacitance value

\[
C = k\epsilon_0 \frac{A}{d}
\]  

(8)

at no external pressure for $C_1$, $C_2$, and $C_3$ is shown in Table 4. The COMSOL capacitance column shows the capacitance of one hexagonal module (that is 6 times the calculated capacitance from the triangular geometry shown in Figure 27), and it completely matches the analytical results.

Table 4. Partial capacitance at no external pressure by analytical calculations

<table>
<thead>
<tr>
<th>Dielectric</th>
<th>Area</th>
<th>Gap</th>
<th>Capacitance (pF)</th>
<th>Analytical</th>
<th>COMSOL</th>
</tr>
</thead>
<tbody>
<tr>
<td>$C_1$</td>
<td>PDMS/Parylene-C</td>
<td>$A_s$</td>
<td>$d_{sp} + d_{pr}$</td>
<td>0.001</td>
<td>0.001</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$313 \times A_s$</td>
<td></td>
<td>0.4</td>
<td>-</td>
</tr>
<tr>
<td>$C_{1t}$</td>
<td>Air</td>
<td>$A_m - A_s$</td>
<td>$d_g = 5 \mu\text{m}$</td>
<td>0.552</td>
<td>0.552</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$A_t - 313 \times A_s$</td>
<td></td>
<td>176.9</td>
<td>-</td>
</tr>
<tr>
<td>$C_2$</td>
<td>Parylene-C</td>
<td>$A_m - A_s$</td>
<td>$d_{pr}$</td>
<td>13.482</td>
<td>13.482</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$A_t - 313 \times A_s$</td>
<td></td>
<td>4324.5</td>
<td>-</td>
</tr>
<tr>
<td>$C_{3t}$</td>
<td>PDMS</td>
<td>$A_m - A_s$</td>
<td>$d_{pd}$</td>
<td>6.320</td>
<td>6.320</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$A_t - 313 \times A_s$</td>
<td></td>
<td>2027.1</td>
<td>-</td>
</tr>
</tbody>
</table>
According to the connection of subcapacitors in Figure 31, and analytically calculated values in Table 4, the total capacitance of a single sensing module

\[ C_{\text{module}} = C_1 + \frac{1}{\frac{1}{C_2} + \frac{1}{C_3} + \frac{1}{C_4}} \]  

(9)

is calculated as 0.490 pF and is equal to the numerically calculated value by COMSOL. In the same way, the total capacitance of the sensor

\[ C_{\text{total}} = C_{1t} + \frac{1}{\frac{1}{C_{2t}} + \frac{1}{C_{3t}} + \frac{1}{C_{4t}}} \]  

(10)

is calculated as 157.2 pF. Based on equation 9, the sensitivity of the sensor to pressure can be calculated as:

\[ \frac{dC}{dP} = \frac{1}{C_2^2 \left( \frac{1}{C_2} + \frac{1}{C_3} + \frac{1}{C_4} \right)} \frac{dC_2}{dP} \]

assuming that only the capacitance of the air gap \( C_2 \) is a function of pressure, \( P \). It explains why we only considered the height and spacing of the spacers as our optimization parameters, because these two parameters set the dimensions and magnitude of \( C_2 \). However, the sensor can have a higher sensitivity for greater values of \( C_3 \) and \( C_4 \). Due to the geometry of the sensor and also fabrication-related constraints, the only independent parameter of \( C_3 \) and \( C_4 \) is their thickness. In section 2.2.2, it is explained how we managed to minimize the thickness of the PDMS layer (hence increase \( C_4 \)) to maximize the sensitivity of the sensor.

3.3.2 Wrinkles

In reality, after depositing the metal layers on the PDMS membrane, due to a big difference in thermal coefficient of expansion of PDMS and metals, the metal layer buckles and forms random 2D wrinkles with a regular wavelength and amplitude. The average wavelength and amplitude of the wrinkles are measured using the Profilm3D profilometer as 2.6 ± 0.2 μm and 0.25 μm, respectively (see section 2.1).

The simulations of the membrane with flat metal layer show that the metal layer is what makes the membrane stiff. To study the effect of the wrinkled gold layer, the deflection of two membranes under
30 Pa is studied using a COMSOL simulation. The two membranes have a similar radius, layer thickness, and boundary condition, except in one the gold layer is flat and in the other it is wrinkled. The wrinkles in the gold layer are implemented in the simulation as concentric corrugations; a schematic of this axisymmetric membrane geometry is depicted in Figure 32. It is expected that this simplification leads to a softer membrane compared to one with a random arrangement of wrinkles such that the compliance of a membrane with random wrinkles will be somewhere between the compliance of the flat membrane and the one with concentric corrugations. Eventually, one can conclude that a wrinkly metal layer, decreases the membrane’s stiffness. It results in higher deflection for same pressure and membrane dimensions, which means higher sensitivity.

![Figure 32](image.png)

*Figure 32. The geometry of the axisymmetric membrane with wrinkled gold layer with fixed and roller boundary conditions at the edge of the spacer (at position 10 μm) and end of the membrane, respectively*

The maximum deflection of the membrane at its edge is reported in Table 5. As it suggests, the wrinkly membrane is softer than the membrane without wrinkles, which, so far, we based all the design on. It is expected that the maximum deflection of a membrane with random wrinkles is somewhere between 4 and 14 μm. This softer membrane will increase the sensitivity of the sensor to pressure changes, and it is to the benefit of the sensor.
Table 5. Effect of wrinkled gold layer on maximum deflection results of simulation

<table>
<thead>
<tr>
<th>Membrane geometry</th>
<th>Max. deflection (μm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Without wrinkle (flat gold layer)</td>
<td>4.0</td>
</tr>
<tr>
<td>With wrinkle</td>
<td>14.8</td>
</tr>
</tbody>
</table>

The difference between the results of the flat and the wrinkled metal layer is significant. Since the Young’s modulus of PDMS is negligible compared to the gold, the metal electrode has the main role in the deflection of the membrane. By changing the geometry of this layer from a flat to a wrinkly layer, the length increases. Since the distribution of pressure is similar for both the flat and the wrinkly membrane, the radial stress as a function of radial position is similar as well. This stress will tend to flatten the wrinkles and lead to a larger radial elongation on this layer that is associated with more membrane deflection.

The first resonance frequency of the wrinkled model with the boundary conditions shown in Figure 32 is calculated in COMSOL as 1116 Hz. The sensor is designed to work for below 1 Hz working condition, therefore, this design is safe.
4.1 Membrane deflection test

In this experiment, the goal is to find the correlation of the membrane deflection to pressure. For that purpose, a setup is made that can control the pressure inside the cavity of the sense capacitor of the sensor (the inner side of the membrane), and leaves the outside of the membrane open to the microscope (Profilm3D, Filmetrics Inc.). Pressure is produced by a 25 μl syringe with luer-lock fitting and monitored by a reference pressure sensor (HSCDRRN001ND2A3, Honeywell). Figure 33 shows the components and their connections of this test setup. To distinguish our proposed sensor from the off-the-shelf pressure sensors, it is called SPX1.

![Figure 33. Pressure-Deflection test setup components and their connections](image)

The setup has a stand to keep SPX1 horizontally flat under the microscope as shown in Figure 34. This stand was modelled in Solidworks and made by 3d printing (Play, Printrbot). It has a path for 1/8” tubing to be connected to the PDMS chamber. The tubing is fitted into the opening of the chamber with
a diameter of 1 mm. The chamber is made by casting PDMS in a 3d printed mold modelled in Solidworks. Vacuum grease is used for sealing together with a zip-tie around the neck of the chamber. To prevent the sharp edge of the zip-tie from cutting the chamber neck, a protective soft layer (e.g. rubber ribbon, Teflon tape etc.) is wrapped around the chamber neck. The SPX1 sensor is placed on the chamber in a way that the electrodes face the inner side of the chamber. The top clamp is fixed on the sensor using four M5 screws.

![Diagram of the sensor setup](image)

*Figure 34. Pressure-Deflection test setup*

The plan is to scan the movement of the membrane between three spacers due to pressure change in the cavity of the sensor. However this experiment failed for two reasons:

1) The membrane was not flat and it was hard to scan it using the optical profilometer.
2) The sensor itself deflects as the chamber is pressurized, not the cavity directly. With the optical profilometer, interference fringes were visible, that were moving due to pressure changes, but they indicated rigid body motion of the whole sensor, not just deflection of the membrane.

4.2 Membrane vibration

The alternate experiment to visualize the movement of the membrane, was to use a Laser Doppler Vibrometer (LDV) (MSA-500, Polytec) while the electrodes of the capacitive sensor are excited with a signal generator. The excitation signal was a DC bias of 2 V added to a 0.5 V AC signal containing the frequency range of 100 Hz to 2 kHz. The signal was amplified 20 times by a signal amplifier (Precision amplifier 2350, Tegam Inc.). The results of LDV are shown in Figure 35 and the distance between the peaks of motion matches the spacing between the modules. Figure 36 shows the frequency spectrum of the
membrane. The vertical axis is filtered by a moving average filter with a span of 3 and is normalized to the amplitude of the resonant peak.

![Grid measurement of the membrane using LDV](image)

Figure 35. Grid measurement of the membrane using LDV

![Vibration spectrum of a vibrating point on the membrane measured by LDV](image)

Figure 36. Vibration spectrum of a vibrating point on the membrane measured by LDV

The experimental resonance frequency, i.e. 1155 Hz, matches the simulation result of the simplified geometry with wrinkled membrane, 1116 Hz (see section 3.3.2).

4.3 Pressure-capacitance relationship

In this experiment, the performance of the sensor is studied, in particular the change in capacitance caused by a change in pressure. Figure 37 shows the tube and wire connections of the pressure-
capacitance test setup. An LCR meter (HM8118 programmable LCR-Bridge, Rohde & Schwarz) is used to measure the capacitance of the sensor. The data was recorded on the laptop by a free control program (by Kaktus circuits, Version 1.1.2.0) or using Python serial communication coding. The LCR meter is calibrated before each measurement session. To have the least amount of parasitic capacitance, the measurements are done at a frequency of 20 Hz which is the lowest frequency that this LCR meter can measure. The pressure is controlled by a 25 μl syringe and is monitored by a reference pressure sensor (HSCDRRN001ND2A3, Honeywell Solutions), read by an Arduino. The fixture is shown in Figure 38.

Figure 37. Pressure-capacitance test setup components and their connections
As shown in Figure 39, in a cycle through the pressure range from −50 to 100 Pa, hysteresis is observed. The rising and falling curves are averaged for each pressure value to derive the characteristic curve of the sensor. As expected it has a different behavior at negative pressure values compared to the positive pressure range (see section 3.2). Using the best fit line for this pressure range, yields the absolute sensitivity of the sensor 0.52 pF/Pa and a relative sensitivity of 0.002 Pa⁻¹ or 2.064 kPa⁻¹ with a zero offset of 252 pF. The maximum hysteresis and non-linearity errors are 11.6%FS and 11.0%FS, respectively. For a more accurate sensor read out, one can use a look up table.
Figure 39. Sensor transfer behavior measured over a full cycle

The experimental results show that SPX1 has a capacitance of 252 pF at absence of external pressure that is higher than the calculated value of 157 pF in section 3.3.1; a reason for this is probably that the membrane is not completely stretched between spacers as assumed when modeled in COMSOL and as assumed for the analytical calculations. The direct effect of a loose membrane is a shorter gap for the air-filled part of the cavity, C2 and C2t (introduced in section 3.3.1). By recalculating the capacitance results with only changing the air-filled cavity height, \( d_g \), from 5 to 2.87 μm based on trial and error, the analytical and experimental values become closer, as shown in Table 6.

<table>
<thead>
<tr>
<th></th>
<th>Capacitance (pF)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Analytical</td>
<td>Experimental</td>
</tr>
<tr>
<td>1 Module</td>
<td>0.726</td>
</tr>
<tr>
<td>Total</td>
<td>252.317</td>
</tr>
</tbody>
</table>

In conclusion, it appears that the initial gap length is smaller than 5 μm due to the loose membrane, thus, the membrane touches the lower layer at a lower pressure (than what was expected based on the flat membrane design) and with increase of pressure, a larger area of the membrane lies on the lower layer. However, it does not stick to the lower layer, because of the Parylene-C coating.
As mentioned the fabrication chapter, the Parylene-C coating prevents stiction between metal electrode and the spacers. It causes the membrane to deflect differently for positive and negative pressure as shown in Figure 40. For negative pressures below -10 Pa, the membrane is separated from the spacers and is deflected as a large square membrane. For pressures between -10 Pa and 20 Pa, the membrane is suspended between the spacers and deflects as hexagonal modules. This regime has the highest sensitivity. For pressures above 20 Pa, the expanded membrane touches the parylene-C coated opposite surface of the cavity and the area of contact slowly increases as the pressure is increased.

![Figure 40](image)

Figure 40. (a) Deflection of the whole membrane at negative pressure (b) Modular deflection of the membrane at positive pressure (c) Touch mode regime of the modular deflection in high positive pressure, shown for only one module.

To better characterize the three distinct behavior of the sensor in its pressure range, a 3rd order polynomial is used to fit the pressure as a function of capacitance, that results in an absolute sensitivity of 0.864 pF/Pa around atmospheric pressure and maximum calibration and hysteresis errors of 2.7 and 7.3 %FS, respectively.

To measure the noise level of SPX1 output, the sensor, clamped in its fixture, is connected to the LCR meter with no pressure source. Using over 450 data points at no external pressure, the standard deviation (STD) was calculated as 0.097 pF, which can be converted to 0.56 Pa using the linear calibration curve. We can introduce the resolution of SPX1 sensor as 3 times the STD as 1.7 Pa. The resolution of the capacitance measurement of the LCR meter is 1 fF which is 2 orders of magnitude smaller than the calculated STD.
4.4 Long term measurement

For the long term measurement test, a pressure control system (OB1 MK3+ microfluidic flow controller, Elveflow) is used as the pressure controller. The smallest increment of the pressure source was 0.3 PSI (2 kPa) which is 20 times higher than the nominal range of the sensor. However, due to limited access to equipment due to COVID-19 safety protocols, this is the only programmable pressure source we were able to access to generate long pressure cycles. As shown in Figure 41, the pressure controller is set to generate near square waves with a period of 10 seconds and pressures between zero and 2 kPa gauge pressure with some overshoot at the rising edge. This signal is measured by a reference pressure sensor (HSCDANN001PDAA3, Honeywell) used instead of the proposed sensor, in the same setup.

![Figure 41. A single pressure wave generated by the pressure controller measured by a reference pressure sensor](image)

The results of the long-term measurement after about 200 cycles of overpressure is shown in Figure 42.
Exposing the pressure sensor under test to this high pressure represents simultaneously an overpressure test. To test the long-term effect of overpressure, three sensor calibrations are performed, right before, right after, and 3 hours after the pressure cycle, as shown in Figure 43.
According to Figure 42, the capacitance signal drifts toward higher values after around 200 s. This upward drift is also visible in the calibration curve recorded right after the test in Figure 43. This might be due to a temporary adhesion of the membrane to the lower layer as a result of significant over pressure. However, this drift disappeared after 3 hours (or less) and the calibration curve returned back to the original curve that was measured before the test. It proves that SPX1 has a stable performance on long term measurements even under harsh circumstances.

4.5 Orientation dependency

In this test, the goal is to study if the orientation of the sensor has any significant effect on its output. The fixture shown in Figure 38, is taped to a board along with the alligator clips of the LCR meter, so that the whole setup rotates as a rigid body. Since there is no pressure source, the sensor is at zero gauge pressure. As shown in Figure 44, the setup is rotated every 10 seconds, but the output of the sensor does not change drastically. The setup orientation at zero, 90, and 180 degrees is horizontal facing up, vertical, and horizontal facing down, respectively. The dotted parts of the curve, are manually eliminated as they are due to movement of the alligator clips.
4.6 Frequency response

To evaluate the frequency response of SPX1, both the sensor and the reference sensor (HSCDRRN001ND2A3, Honeywell Solutions) are simultaneously exposed to the same pressure signal that includes frequent pressure fluctuations with frequency content over the frequency range of interest. The excitation pressure is generated by randomly or systematically changing the pressure using the 25 μl syringe in the setup shown in Figure 37. The reference sensor has a response time of better than 0.45 ms. The capacitance of the sensor under test is recorded over time and it is then converted to a pressure signal using the calibration equation of the sensor. Figure 45 shows the signals of the two sensors.
The time domain signals are around 600 seconds which limits the low range of the frequency spectrum to 0.0016 Hz. The high end of the frequency spectrum is limited by the sampling rate. We sampled the data from the LCR meter every 0.13 seconds, which is 7.69 Hz. The sampling rate from the reference pressure sensor with Arduino was done at a rate of 10 Hz. To be able to get the Fast Fourier Transform (FFT) of both signals at the same frequencies, the time signals are oversampled at 100 Hz, however the output of the FFT is only considered in the range 0.1 to 5 Hz to prevent numerical artifacts. The FFT of the converted response of SPX1 and the reference sensor are shown in Figure 46, along with a signal smoothed using a moving average with span of 100. Both FFTs show significant frequency content from 0.1 Hz and 5 Hz, which well includes the frequency range of interest 0.2 to 0.5 Hz, since normal respiration rate is between 12 to 28 breathe per minute [54].

*Figure 45. Time domain pressure signal of SPX1 and the reference pressure sensor for evaluating dynamic response of SPX1*
The smoothened FFT of the sensor under test is divided by the smoothed FFT of the reference sensor signal. As shown in Figure 47. The dynamic response of the sensor does not decrease significantly at high frequencies and it stays nearly equal to 1 over the frequency range 0.1 – 5 Hz. However, it has significant fluctuations that indicate artifacts in the measurement. Figure 47 also shows that the fluctuations are not a results of calibration error, since both linear and 3rd order polynomial fits have very similar behavior. Figure 46 shows that both FFTs include minor numerical artifacts in the form of periodic signal features, and the fluctuations in the signal in Figure 47 are mainly caused by a combination of these signal features with different periodicity.
To measure the response time of SPX1 sensor, we tried to make square wave excitation by the 25 μl syringe. The sampling rate via the serial communication (every 0.13 seconds) between laptop and LCR meter was faster than the actual sampling rate of the LCR meter. That is reason we see some points in the SPX1 curve in Figure 48, are having the exact same values; we just recorded the same measurement twice before the LCR meter had the chance to get the next measurement from the sensor. It not only makes problem, finding the actual response time of the sensor, but also would have had some effect on the frequency response of the sensor. At the end, with the current measurement, we can claim that the response time of the sensor is faster than 0.26 seconds.
4.7 Application to the measurement of breathing

As a proof for the capability of the sensor for diagnosing apnea and hypopnea sleep disorders, human respiration (author, Hamed Pouriyayevali) is measured by the SPX1 sensor. The test setup is the same as the one shown in Figure 49, except instead of the syringe, the tube is placed on the upper lip, 1 cm away from the nostril, along the nasal respiration flow. The reference pressure sensor is a HSCDRRN001ND2A3, Honeywell Solutions.

Figure 49 shows the results of this test, where normal, and reduced-depth respiration (as hypopnea, from second 80 to second 90), and total cessation of breathing (as apnea, from second 115 to second 125) are demonstrated. It shows that SPX1 has an adequate performance for respiration monitoring and a sleep professional can use its recorded data for sleep apnea diagnosis.

![Figure 49. Real breathing test of the sensor](image-url)
5.1 Conclusions

The focus of this study was to propose a more comfortable solution for respiration monitoring for the purpose of sleep apnea diagnosis. We designed, fabricated, and characterized a novel modular PDMS-based capacitive pressure sensor. We chose capacitive pressure sensing over other candidate technologies for respiration monitoring for its advantages such as accuracy, mechanical robustness, and power consumption. The sensor is made using several hexagonal capacitive sensing modules that are connected to each other in parallel. This modular design is optimized to not only have a high absolute or relative sensitivity of 0.52 pF/Pa or 2.064 kPa$^{-1}$, respectively, but also a high capacitance of 252 pF, with very low noise level for the pressure range of -50 to 100 Pa. The sensor could undergo overpressure without permanent damage or shift in its performance. The relative sensitivity of SPX1 comes from the module design and its total capacitance comes from the cluster of modules connected in parallel. Therefore, by changing the total number of modules, the final size of the sensor can be controlled, and the relative sensitivity would be the same, however, the total capacitance and SNR would change accordingly. This work can easily stand out among other designs we discussed in the Introduction chapter, as shown in Table 7.
The fabrication steps involve relatively inexpensive techniques and do not need to be done in a cleanroom environment. This allows for a low cost of the sensor, and allows fabrication in any part of the world with no need for access to expensive equipment, except the sputter coater. The materials used for the fabrication are PDMS, SU-8 2005, carbon black powder, Parylene-C, titanium, and gold. We used a spin coater, a planetary centrifugal mixer, a plasma cleaner, a mask aligner, a Parylene-C coater, and a sputter deposition machine for the fabrication steps.

An LCR meter is used to read the capacitance of the sensor through an interface software on a laptop. Two test setups were 3d-printed for the pressure-deflection, and pressure-capacitance tests. We were unable to observe the movement of the membrane during operation, however, the capacitance results are in good agreement with the theoretical expectations of the sensor. The response time of the

---

<table>
<thead>
<tr>
<th>Membrane Material</th>
<th>Cui et al. [48]</th>
<th>Ma et al. [51]</th>
<th>Pedersen et al. [47]</th>
<th>Berger et al. [49]</th>
<th>This Work (SPX1)</th>
</tr>
</thead>
<tbody>
<tr>
<td>PDMS</td>
<td>PDMS</td>
<td>Silicon</td>
<td>Graphene</td>
<td>PDMS</td>
<td></td>
</tr>
<tr>
<td>Diameter/Diagonal (Module’s)</td>
<td>1 cm</td>
<td>1 cm</td>
<td>2.3 mm (150 μm)</td>
<td>210 μm (30 μm)</td>
<td>1 cm (600 μm)</td>
</tr>
<tr>
<td>Gap Width</td>
<td>50 μm</td>
<td>-</td>
<td>420 nm</td>
<td>180 nm</td>
<td>5 μm</td>
</tr>
<tr>
<td>Relative Sensitivity (Absolute)</td>
<td>0.741 kPa⁻¹</td>
<td>2.04 kPa⁻¹</td>
<td>(0.76 pF/kPa)</td>
<td>(0.004 pF/kPa)</td>
<td>2.064 kPa⁻¹ (0.52 pF/Pa)</td>
</tr>
<tr>
<td>Range</td>
<td>0 – 1 kPa</td>
<td>0 – 2 kPa</td>
<td>200 – 600 kPa</td>
<td>0 – 100 kPa</td>
<td>-50 – 100 Pa</td>
</tr>
<tr>
<td>Initial Capacitance</td>
<td>3.5 pF</td>
<td>4.5 pF</td>
<td>106 pF</td>
<td>137.6</td>
<td>252</td>
</tr>
</tbody>
</table>
sensor is shorter than the sampling interval of the logging platform, 0.86 seconds, and it is sufficient to capture apneic events in human respiration during sleep.
### Table 8. SPX1 assessment table

<table>
<thead>
<tr>
<th>#</th>
<th>Requirements</th>
<th>Pass/ Inconclusive</th>
<th>Notes</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Smaller than 1 cm²</td>
<td>Pass</td>
<td>The design is applicable for any size of the sensor.</td>
</tr>
<tr>
<td>2</td>
<td>Flexible</td>
<td>Inconclusive</td>
<td>The sensor is made of flexible material and can undergo certain bending angles without permanent damage.</td>
</tr>
<tr>
<td>3</td>
<td>Error less than 15%</td>
<td>Pass</td>
<td>The hysteresis and nonlinearity errors are 11.6%FS and 11.0%FS, respectively.</td>
</tr>
<tr>
<td>4</td>
<td>Resolution smaller than 3.5 Pa</td>
<td>Pass</td>
<td>The resolution is 1.7 Pa.</td>
</tr>
<tr>
<td>5</td>
<td>Cover -50 to 100 Pa pressure range</td>
<td>Pass</td>
<td>The sensor successfully measures the pressure in this range with high sensitivity.</td>
</tr>
<tr>
<td>6</td>
<td>Tolerate overpressure</td>
<td>Pass</td>
<td>The sensor did not have any permanent damage after 200 cycles of 20x overpressure at 2 kPa.</td>
</tr>
<tr>
<td>7</td>
<td>Response time less than 0.1 s</td>
<td>Inconclusive</td>
<td>The sampling frequency is too low to measure any response time less than 0.25 seconds.</td>
</tr>
</tbody>
</table>

#### 5.2 Future Work

In this work, a prototype and a proof of concept of a modular PDMS-based capacitive pressure sensor was successfully demonstrated. However, we are far from having an integrated Type IV respiration monitoring device.

The loose membrane is by far the biggest problem of this work. There is a need for studying the initial deflection of the membrane to be able to define all the effective parameters of the module design (as opposed to only optimizing the spacer height and pitch). The final size of the sensor is still too large for mounting on the upper lip of the patient. It can be minimized by moving the connections on the CB electrode, rather than 5 mm away from it. The size of the electrodes can also be minimized with cost of having lower absolute sensitivity. The other aspect that this work could have been improved in, is the long
term measurement. The sensor needs to be tested for a long term measurement of at least 10000 full cycles (over 8 hours) within its range of operation to prove its ability to monitor respiration for a full night. The bending tolerance of the sensor also has to be investigated if it is going to be implemented in a flexible face-borne device.

To have an integrated device, as explained in section 1.4, we can proceed with different methods of data storage and communication. For the impedance-loaded SAW method, a multi-reflector SAW delay line is needed with proper antenna. The sensor must be connected to one the reflectors after matching the impedances of the sensor and the reflector. An impedance analyzer will be needed to study the performance of the data communication between the transmitter antenna and the SAW-based device. An alternative way to using SAW, is to read the capacitance value by a capacitance-to-digital converter such as FDC2112DNTT, Texas Instruments. A Bluetooth unit can then be used to transmit data to a data collection unit.
References


[32] A. Al-Salaymeh, J. Jovanović, and F. Durst, “Bi-directional flow sensor with a wide dynamic range


material based on rigid low-melting-point-alloy-microstructures embedded in soft poly(dimethylsiloxane) (PDMS),” 2013.


[77] L. Reindl and W. Ruile, “Programmable reflectors for SAW-ID-tags,” in *Proceedings of the IEEE* 66

Appendices

Appendix 1: SAW

Piezoelectric materials are solid materials such as certain crystals or ceramics that can accumulate electric charge on surface electrodes in response to a mechanical stress and this effect is reversible. Interdigital transducers (IDTs) are pairs of periodic metal stripes; when deposited on a piezoelectric substrate they can allow exciting and detecting SAWs. IDTs were first introduced by White and Voltmer in 1965 [62]. Given a piezoelectric substrate along with the appropriate IDT geometry, a voltage waveform applied across the IDT terminals will result in a SAW propagating away from the IDT array towards the sensing IDT array. When a transmitted SAW reaches the receiver IDT, the mechanical wave can be converted back to an electrical waveform. Figure 50 shows some examples of IDT arrays commonly used in SAW devices. The geometry of the IDT defines the generated SAW. For example in a single-electrode type IDT, shown in Figure 50.a, the pitch (p) of the fingers is equal to the wavelength (λ) of the SAW it generates.

![Figure 50. a. A single-electrode type IDT. b. A double-electrode type IDT. c. A short-circuited grating. d. An electrode-width-controlled single phase unidirectional transducer (EWC/SPUDT)](image)

There are several methods for modeling IDTs. Here, the p-matrix method, shown below, is briefly discussed as we need it to explain how some SAW sensors work. The p-matrix models the IDT as a black box with \( a_1, a_2, \) and \( V \) as inputs, and \( b_1, b_2, \) and \( I \) as outputs, where \( a_1, \) and \( a_2 \) are the SAWs that traveled toward the IDT from its left and right, respectively, \( b_1, \) and \( b_2 \) are the reflected SAWs from or generated...
SAWs by the IDT toward its left and right, respectively, \( V \) is the voltage between IDT pair, and \( I \) is the current passing through the IDT. Figure 51 shows the inputs and outputs of the IDT black box model for defining the \( p \) matrix.

\[
\begin{bmatrix}
  b_1 \\
  b_2 \\
  I
\end{bmatrix}
= \begin{bmatrix}
  p_{11} & p_{12} & p_{13} \\
  p_{21} & p_{22} & p_{23} \\
  p_{31} & p_{32} & p_{33}
\end{bmatrix}
\begin{bmatrix}
  a_1 \\
  a_2 \\
  V
\end{bmatrix}
\]

*Figure 51. Inputs and outputs to IDT black box*

To illustrate how a SAW device and a pressure sensor might be used in a sensor-transmitter-receiver system it is useful to refer to the seminal work by Reeder et al. [63]. Reeder et al. used SAW resonators to make a passive wireless pressure and temperature sensor. The SAW resonator was made on the thin diaphragm that could deflect by pressure and change the resonance frequency of the resonator. There are many passive sensors that have used different SAW technologies [64]–[69]. In these works, the membrane is part of the SAW sensor which brings some limitations in thickness and stiffness of the membrane for a low-range pressure sensor. To overcome this issue, there is another group of SAW sensors. As shown in Figure 52, in this group, an external sensor is used for sensing and SAW is used to communicate with the sensor passively and wirelessly [70]–[74]. In appendix, a similar external sensor-loaded SAW device is explained.
The amplitude of the reflected pulse from the impedance-loaded IDT (shown as target reflected pulse in Figure 52) is compared to a reference to compensate for the energy dissipations of the environment and other components. With the same principle, there can be more impedance-loaded IDTs on a single substrate each connected to a separate sensor (e.g. a pressure sensor, a humidity sensor, and a temperature sensor) and then the reflected waves will be compared to the reference one separately.

Due to pressure change, the impedance of the pressure sensor changes that affects the reflected SAW by the loaded IDT through the acoustic reflection coefficient of the loaded IDT

$$p_{11}(Z_{\text{load}}) = p_{11}^{\text{SC}} + \frac{2p_{13}^2}{p_{33} + \frac{1}{Z_{\text{load}}}}$$

where $p_{11}^{\text{SC}}$ is the acoustic reflection coefficient of the short-circuited IDT, $p_{13}$ is the electroacoustic coupling factor, and $p_{33}$ is the electrical admittance of the IDT [75]–[78]. $Z_{\text{load}}$ is the combined impedance of the sensor and the impedance-matching elements. An increase in $p_{11}$ of the loaded IDT increases the amplitude of the reflected SAW.

A sensor with higher sensitivity (larger relative impedance change) with respect to pressure is preferred. When the magnitude of the impedance of a capacitive sensor

$$|Z|_{\text{capacitor}} = \frac{1}{\omega C}$$

with $\omega$ as the frequency of the SAW device, is derived with respect to $C$, this yields

Figure 52. Schematic of a wireless passive sensor based on SAW technology
Whereas the impedance of a resistive sensor

\[ |Z|_{\text{Resistor}} = R \]  \hspace{1cm} (14)

has a lower sensitivity compared to a capacitive sensor with a capacitance in Pico farad range and \( \omega \) in Megahertz range \( \frac{1}{\omega C^2} \gg 1 \). Considering equation 11, the sensitivity of the SAW device to pressure

\[
\frac{\partial p_{11}}{\partial P} = \left( \frac{\partial p_{11}}{\partial Z_{\text{load}}} \right) \left( \frac{\partial Z}{\partial X} \right) \left( \frac{\partial X}{\partial P} \right)
\]

is derived, where \( P \) is the pressure and \( X \) is either the resistance (\( R \)) or the capacitance (\( C \)) of the sensor. So a capacitive sensor can be a better candidate for implementing in a passive impedance-loaded SAW device.