İrfan SOLAK A Master's Thesis AGU 2022

RESPIRATION MONITORING USING A FLEXIBLE PAPER-BASED CAPACITIVE SENSOR

A THESIS
SUBMITTED TO THE DEPARTMENT OF ELECTRICAL AND COMPUTER ENGINEERING
AND THE GRADUATE SCHOOL OF ENGINEERING AND SCIENCE
OF ABDULLAH GUL UNIVERSITY
IN PARTIAL FULFILLMENT OF THE REQUIREMENTS
FOR THE DEGREE OF
MASTER OF SCIENCE

By
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SCIENTIFIC ETHICS COMPLIANCE

I hereby declare that all information in this document has been obtained in accordance with academic rules and ethical conduct. I also declare that, as required by these rules and conduct, I have fully cited and referenced all materials and results that are not original to this work.

All human samples in this study were obtained under the approval of the Research Ethics Committee of the Abdullah Gül University (Approval date: 25.04.2022, Decision no:33079, Kayseri, Turkey). Written informed consent was obtained from all individuals. The Declaration of Helsinki was followed throughout the study.

All patient samples in this study were obtained under the approval of the Clinical Research Ethics Committee of the Kayseri City Hospital (Approval date: 12.01.2022, Decision no:58, Kayseri, Turkey).

Name-Surname: İrfan SOLAK
Signature :

REGULATORY COMPLIANCE

M.Sc. thesis titled Respiration Monitoring Using a Flexible Paper-Based Capacitive Sensor has been prepared in accordance with the Thesis Writing Guidelines of the Abdullah Gül University, Graduate School of Engineering & Science.

Prepared By	Advisor	Co-Advisor
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ACCEPTANCE AND APPROVAL

M.Sc. thesis titled Respiration Monitoring Using a Flexible Paper-Based Capacitive Sensor and prepared by İrfan SOLAK has been accepted by the jury in the Electrical and Computer Engineering Graduate Program at Abdullah Gül University, Graduate School of Engineering & Science.

27/07/2022

(Thesis Defense Exam Date)

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APPROVAL:

The acceptance of this M.Sc. thesis has been approved by the decision of the Abdullah Gül University, Graduate School of Engineering & Science, Executive Board dated ….. /….. / ……….. and numbered ……………..……. .

……….. /……….. / ………..

(Date)

Graduate School Dean
Prof. Dr. İrfan ALAN

ABSTRACT

RESPIRATION MONITORING USING A FLEXIBLE PAPER-BASED CAPACITIVE SENSOR

İrfan SOLAK
MSc. in Electrical and Computer Engineering
Advisor: Assoc. Prof. Kutay İÇÖZ
Co-Advisor: Asst. Prof. Dooyoung HAH

July 2022

Respiration is an action known to be essential and crucial for life. Unfortunately, in some cases such as illnesses and accidents various respiratory problems can be experienced. It might be difficult to maintain normal respiration for the people who have respiratory diseases. It is known that respiration monitoring of people who have respiratory problems, albeit for different reasons, is important in terms of their treatment and maintaining their life quality. Current respiration monitoring systems are expensive and bulky. Many of these systems are only available at hospitals or in laboratories. Low-cost, easy to use and portable respiratory monitoring devices are needed. Having these motivations, we aimed to monitor respiration by designing and producing a paper-based sensor that is easy to manufacture, low-cost, and highly responsive. The sensor, which is the subject of this thesis project, has potential to be used for different purposes such as measuring the humidity in the environment. In this project, we focused on designing a system for people who have respiratory problems by providing respiration monitoring data. In addition, according to the data obtained, we are able to analyze the health status of the users. Therefore, this sensor can be used both for the detection of respiration diseases and monitor the status of the patients. In this way, respiration related unhealthy situation can be detected and treated immediately.

Keywords: Capacitive sensor, Respiration, Electronics, Wearable, Paper-based

ÖZET

ESNEK KÂĞIT TABANLI KAPASİTİF SENSÖR KULLANARAK SOLUNUM İZLEME

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Temmuz-2022

Solunum, yaşam için gerekli ve çok önemli bir eylemdir. Ne yazık ki hastalık ve kaza gibi bazı durumlarda çeşitli solunum problemleri yaşanabilmektedir. Solunum hastalıkları olan kişilerde solunumu sağlamak zor olabilir. Farklı nedenlerle de olsa kişilerin solunumunu takip etmenin onların tedavisi ve yaşam kalitelerinin artması açısından önemli olduğu açıktır. Mevcut solunum izleme sistemleri pahalı ve hantaldır. Bu sistemlerin çoğu sadece hastanelerde veya laboratuvarlarda mevcuttur. Düşük maliyetli, kullanımı kolay ve taşınabilir solunum izleme cihazlarına ihtiyaç vardır. Bu ve benzeri nedenlerle, üretimi kolay, düşük maliyetli ve yüksek duyarlıklı kâğıt tabanlı bir sensör tasarlayıp üreterek solunumu izlemeyi hedefledik. Bu tez projesine konu olan sensör, ortamdaki nemin ölçülmesi gibi farklı amaçlarla kullanılabilmektedir. Bu projede solunum izleme verilerini sağlayarak solunum problemi olan kişiler için bir sistem tasarlamaya odaklandık. Ayrıca elde edilen verilere göre kullanıcıların sağlık durumları hakkında da bilgi alabiliyoruz. Dolayısıyla bu sensör hem sağlık durumunun tespiti hem de hastaların takibi için kullanılabilmektedir. Bu sayede solunum problemleri hemen tespit edilebilecek ve tedavileri hızlı bir şekilde gerçekleştirilecektir.

Anahtar kelimeler: Kapasitif sensor, Solunum, Elektronik, Giyilebilir, Kâğıt tabanlı

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I cannot begin to express my thanks to my precious parent, who I know that they are always by my side, to my mother Tülay SOLAK and my father Yılmaz SOLAK for their unwavering support and motivation throughout this process. I was able to complete this thesis thanks to their incredible support. I was able to continue with the great ideas they gave me every time I struggled with the thesis.

I would like to express my special thanks to my sister Esra SOLAK, who contributed greatly to this work and helped me in all aspects. I would not have been able to progress in this study without her incredible help. She was always there to help me even when she had works to keep herself busy.

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LIST OF ABBREVIATIONS

MEMS	Micro-Electro-Mechanical Systems
ECG	Electrocardiogram
CCD	Charge Coupled Device
PMMA	Poly(Methyl Methacrylate)
pF	PicoFarad
MtM	Maximum to Minimum
s	Second
BPM	Breaths Per Minute

To my family

Chapter 1

Introduction

Continuous breathing is essential for life but in some unfortunate circumstances, breathing may have irregularities [1]. Respiratory monitoring is crucial for people having respiratory problems. Current respiratory monitoring methods are often expensive, cumbersome, and not portable. These lab devices are reliable and sensitive however, people need to go to hospitals or clinics for respiration monitoring and measurements. In addition to air pollution [2], there is an increase in respiratory problems due to diseases such as Covid-19 [3]. As a result, respiration monitoring has become more important than ever. To offer a solution to this problem, we have developed a fast, inexpensive, and highly accurate moisture sensor. The moisture sensor uses the hygroscopic character of the paper that is the ability of paper to adsorb water reversibly from the surrounding environment [4]. The moisture in exhaled human breath is absorbed by the paper-based sensor and sensed as a capacitance change. This change shows itself as an increase in capacitance. By measuring this capacitive change, respiration can be analyzed. The moisture content in the paper will decrease during inhale, and it causes a decrease in capacitance. With this method, respiration monitoring can be achieved with a low cost and highly responsive sensor. The following section introduces the respiration and sensors.

1.1 Respiration

For humans, respiration occurs by the exchange of oxygen-containing air with carbon dioxide-containing air. Oxygen-containing air is taken into the lungs using the mouth and nose. Alveoli in the lungs carry out the gas exchange [5]. The alveoli give out the carbon dioxide gas in the capillaries and deliver oxygen gas to the capillaries. Thus, the oxygen needed by the body is provided and the carbon dioxide that needs to be removed from the body is thrown out [6]. The carbon dioxide is then expelled through the mouth or nose. This entire gas exchange system is called respiration.

Respiration, blood pressure, pulse rate, and body temperature are vital signs that provide information about the health status of people [7]. By interpreting these signs, diseases can be detected early. Interpretation can be made thanks to the known general information about respiration like respiration rate etc. In the table below, the respiration rate, one of the most important signs, is given according to different ages [8].

Table 1.1 Respiration rates vs. ages [8]

Age	**Respiration Rate**
<1 year	30 - 40 in a minute
1 - 2 years	25 - 35 in a minute
2 - 5 years	25 - 30 in a minute
5 - 12 years	20 - 35 in a minute
>12 years	12 - 20 in a minute

Using similar information, it is possible to comment on the health status of people. For example, a rate of more than 27 breaths per minute is one of the strongest indications of cardiac arrest [9]. The diagnosis of some diseases can be made using the respiration rate such as sleep apnea [10]. Apart from the respiratory rate, another known information is the average gas change per minute. In adults, 12 to 20 breaths per minute correspond to approximately 6-8 liters of gas exchange [11]. Using this information, doctors can also evaluate the health status of the people. Measuring the compounds of exhaled breath can provide detection of the respiratory illnesses [12]–[14], air humidity can be also used for the diagnosis of pulmonary disease and monitoring the exhaled breath [15]–[19].

1.2 Sensors

Sensors are devices that respond to physical stimuli in the surrounding environment. These physical stimuli can be temperature, sound, light, pressure [20], magnetism, movement, etc. [21]. Sensors are activated with a stimuli and convert that signal into electrical signals. In this respect, sensors are similar to human sense organs. Our eyes are biological sensors that detect the surrounding light, and similarly, our ear is a biological sensor that can detect sounds. The signals obtained from the sensors are generally processed with integrated circuits so various operations can be performed. By using sensors, we can measure the ambient temperature as well as we can detect planets millions of light-years away [22]. We encounter sensors in almost every part of our daily lives.

From the touch screen of the phones to the buttons that we can increase or decrease the brightness of the lamps, from microphones to fingerprint readers, we have many sensors in our daily lives. Figure 1.1 shows some of these sensors.

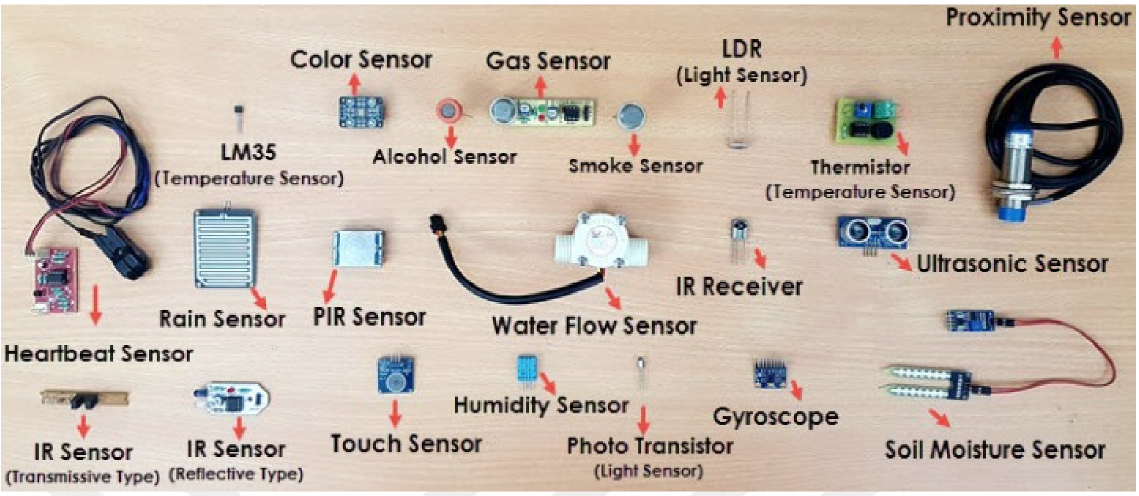

Figure 1.1 Different kinds of sensors [23].

Sensors can be classified into various categories [24]. One of the classifications is active and passive sensors. Active sensors need an external power source, while passive sensors do not need any external power source. Another classification can also be made according to the detection method. Sensors can be classified as biological, electrical, chemical or optical [25], [26]. Classification can be also made according to the output signal. In this case, the sensors are divided into two categories: analog and digital. As the name suggests, digital sensors produce digital output, while analog sensors produce analog output. For example, microscope or cell phone images can be the digital outputs of sensors [27]–[29] and actuators [30], [31]. The number of classifications can be further increased. As technology developed, the sensors began to get smaller and portable [32]. Smaller sensors usually have many advantages, they are generally faster, and more sensitive. Depending on the production method and materials, the cost of sensors may increase or decrease. MEMS technology offers new alternatives to conventional sensors [33]. Micron size sensors can be produced with MEMS technology [34]–[36] and these sensors can be integrated with micro or nano size particles [37] and microfluidic systems [38]–[41]. In addition to being much faster, these sensors can make measurements at a lower cost with higher sensitivity. As a result of the demand for inexpensive high-accuracy sensors, the importance of disposable and point-of care systems has increased in these days [42]. Some sensors require surface functionalization with receptor molecules which would increase the cost of the measurement [43]–[46]. Disposable sensors are accessible to everyone because they are easy to use and inexpensive. The paper-based moisture sensor, which is the subject of this thesis, was also produced with this information in mind. Since it is paper-based, it is low cost and due to the hygroscopic character of the paper, it can be used as a moisture sensor [4], [47], [48]. In addition, other important advantages of the paper sensor are its fast response time and easy use.

Chapter 2

BACKGROUND

Since respiration is one of the most important signs of the health status of people, several studies have been carried out on respiration monitoring from the past to the present. In this section, previous studies on respiration monitoring are briefly explained. The methods used for respiration monitoring are generally divided into two categories contact-based respiration monitoring and noncontact-based respiration monitoring [49], [50]. There are also many subcategories of these two main categories. The sensor mentioned in this thesis can be evaluated as contact-based respiration monitoring since it is a paper-based moisture sensor that can detect the moisture in human breath. Alternatively, respiration monitoring can be done by detecting the temperature [51], as well as by detecting the sound that is generated during the breathing by using a microphone [52]. These two examples are also in the contact-based respiration monitoring category. Radar-based respiration rate monitoring or optical-based respiration rate monitoring can be given as examples of the noncontact-based respiration monitoring category. In radar-based respiration rate monitoring, the chest movement of the subjects is detected using the Doppler phenomenon, so that respiratory monitoring is performed [53]. In optical-based respiration rate monitoring, chest movement is detected with similar logic, but this system is based on optics and generally invisible infrared light is used [54]. A general summary of the systems, which are also given as sub-titles, can be seen below [6].

Contact-Based Respiration Monitoring

- Chest and Abdominal Movement Detection
- Oximetry Probe (SpO2) Based
- Acoustic-Based Methods
- Electrocardiogram (ECG) Derived Respiration Rate
- Airflow-Based Methods
- Transcutaneous CO_2 Monitoring

Noncontact-Based Respiration Monitoring

- Optical-Based Respiration Rate Monitoring
- Thermal-Based Respiration Rate Monitoring
- Radar-Based Respiration Rate Monitoring

2.1 Contact-Based Respiration Monitoring

In this method, as the name suggests, the sensing device is in contact with the user's body or in a position very close to it. If the sensing devices are small enough, the system to be installed with the contact-based respiration monitoring method can be used as a wearable [50]. This allows the user to use this system on a mobile basis without being tied to a place which is one of the biggest advantages of these systems.

2.1.1 Chest and Abdominal Movement Detection

Mercury strain gauges or impedance methods can detect both chest and abdominal movement. Mercury strain gauges are an instrument used to detect changes in the volume of various parts of the body and are used in plethysmography. Similarly, a Plethysmograph is an instrument used to measure a change in volume within an organ or whole body. Two bands are used to measure respiration, the thoracic band, and the abdominal band. The thoracic band wraps around the rib cage, while the abdominal band wraps around the abdomen [55]. These bands are flexible in a way that does not make the user's respiration difficult. Bands are designed to maintain conductivity both when inhaled and exhaled. This system, which responds to the changing body volume during respiration, causes a resistance change. In this way, respiration is detected [56].

2.1.2 Oximetry Probe (SpO2) Based

Unlike other methods, the SpO2 method is used to detect the oxygen rate in the blood [57]. Inhaled oxygen is carried from the lungs to the capillaries and then to the organs by blood. Hemoglobin, which is in the blood and is responsible for carrying oxygen, circulates in the body through the veins. The oximetry probe is used to measure the rate of hemoglobin in the blood using infrared light [58]. Thus, comments can be made about the user's respiration since the oxygen level in the blood is known.

2.1.3 Acoustic-Based Methods

Respiration monitoring can be made using the microphone in acoustic-based methods [59]. In this method, the microphone placed near the throat or mouth-nose detects the respiration sound [60]. The signal is then cleared of noise and only respiration sounds remain. Thus, respiratory monitoring is achieved. The most negative effect of this system is false results due to factors such as speaking, coughing, sneezing, snoring, etc.

2.1.4 Electrocardiogram (ECG) Derived Respiration Rate

For this method, ECG modules are connected to a subject, and ECG is recorded during respiration [61]. Respiration causes fluctuations in ECG. Respiration can be detected by looking at these fluctuations. This method relies on a process known as sinus arrhythmia that is the modulation of ECG. The detection of respiration with this method is called electrocardiogram-derived respiration rate [62]. What makes this method possible is the change in the volume of the lungs during respiration and, accordingly, the change in the position of the heart relative to the electrodes [63].

2.1.5 Airflow-Based Methods

For airflow-based methods, sensors that are placed around the oral or nasal area are used. These sensors detect respiration by taking advantage of the features such as the exhaled breath being warmer, more humid, or containing more CO_2 than the inhaled breath [64]. Temperature sensing [65] using a temperature sensor, moisture sensing [66] using a moisture sensor, and carbon dioxide sensing [67] using a carbon dioxide sensor are among these methods. Some sources also refer to sound detection as an airflow-based method [49]. Wearable tools such as masks, mouthpieces, etc. can be used in these methods.

2.1.6 Transcutaneous CO_2 Monitoring

A heated electrode is applied to the subject’s skin, usually the arm. It serves to monitor the CO_2 exchange in general, rather than constantly monitoring respiration. This method, which is based on the diffusion of gas, is surrounded by a solution that provides to the electrode conductivity. With this method, some respiratory problems can be detected [68]. It is not usually possible to detect the respiration rate with this method.

2.2 Noncontact-Based Respiration Monitoring

The system used for the noncontact-based respiration monitoring method does not need to make physical contact with the subject. In this method, the subject can be comfortable as in daily life, since no tools such as a mask, mouthpiece, etc. are attached to the subject's body. The biggest disadvantage of this method is that the user is tied to a place such as a laboratory where the instruments used in this method are located. Therefore, this system is not suitable for mobile usage [69].

2.2.1 Optical-Based Respiration Rate Monitoring

Respiration monitoring is usually done using infrared light and a camera. Some systems accomplish respiration monitoring by detecting chest movements [54]. In these systems, the interpretation of the obtained image is provided by using various algorithms. The user has to be in the place where the assembly is installed. Although it is ideal for monitoring respiration during sleep, subjects should wait for a long time in an inactive spot during times other than sleep. In addition, it is also available for respiration monitoring using a CCD camera with a filter next to it. Infrared light produces bright spots, and the CCD camera detects these spots. The movement of these bright spots is compared to the previous frame, in this way respiration monitoring is performed [70].

2.2.2 Thermal-Based Respiration Monitoring

Thermal sensors or thermal imaging can be used in this method, which is based on the fact that, in most cases, the exhaled breath is warmer than the inhaled breath. In methods of respiration monitoring using thermal sensors, the sensor is generally in a position close to the user. In this way, the exhaled hot breath is easily detected using the sensor [71]. Then, the collected data is analyzed, and respiration monitoring is performed. Although it is a noncontact-based method, it limits the user's movement since the sensor needs to be near the user. On the other hand, infrared sensors can be used for thermal imaging. With the same logic, respiration monitoring is carried out by detecting the inhaled hot air. Using various algorithms, errors are eliminated, and a more reliable system is obtained. Respiration monitoring can also be done using thermal cameras instead of infrared sensors. Temperature changes are measured around the nose and mouth with thermal cameras [72]. Again, using postprocessing algorithms, it is determined that the hot air corresponds to exhalation, and thus respiratory monitoring is performed. As with the other methods in the noncontact-based respiration monitoring section, in this method, the user has to be in a place where measurement devices were placed.

2.2.3 Radar-Based Respiration Rate Monitoring

Movements of the chest are detected using the Doppler phenomenon. The system operator aims beams at the subject's chest. With this system, besides respiration monitoring, heart rate monitoring can be performed. With this method, both respiration and heartbeat are obtained in the same data. After that, they use several filters to separate these two data. It can follow respiration from relatively long distances exceeding 10 meters. In addition, this system can be used to measure the performance of athletes [73].

Chapter 3

MATERIALS AND METHODS

The production process of the paper-based moisture sensor is detailed in this section, including the materials required and fabrication process of the sensor is explained. How different types of pencils, papers, and stencils affect the sensor, as well as the various materials used as a conductor and their consequences are also discussed in this section. Methods and tools such as laser printers, cutting of PMMA, etc. which were used in the production of stencils are also mentioned. After explaining the optimization of the physical parameters of the sensor, information about the selection of the microcontroller is provided. In order to obtain meaningful results from the sensor, we prepared codes for the Microcontroller which can be found in the appendix section. The 3D head mask, which is designed to make the respiration test more robust, is also reported. After obtaining the simulation results with the COMSOL Multiphysics software, a general cost calculation is included in this chapter.

3.1 Papers

Paper is the most critical material that determines the characteristics of the paper-based moisture sensor. To produce a more sensitive and responsive sensor, the paper should absorb and release humidity. For that reason, a thin and high absorbent paper is desired. Thus, it can detect humidity in respiration quickly, and due to its thinness, it can release humidity to the external environment rapidly, so that measurements can be made without any delay. Papers with very different properties such as thickness, quality, durability, etc. was tested for producing the sensors. It was recorded how these papers responded to humidity when the same patterns drawn on them. At the end of this experimental study, we determined that the best paper among the others is ISOLAB weighing paper. This paper, which is thin enough to quickly absorb the surrounding moisture, is mainly produced in accordance with the weighing process. They have been developed in order not to impair the purity of the weighed materials during and after weighing. They are sold in the form of notebooks of 250 pieces, cut into sizes 100 x 100 mm. It can be seen in the figure below.

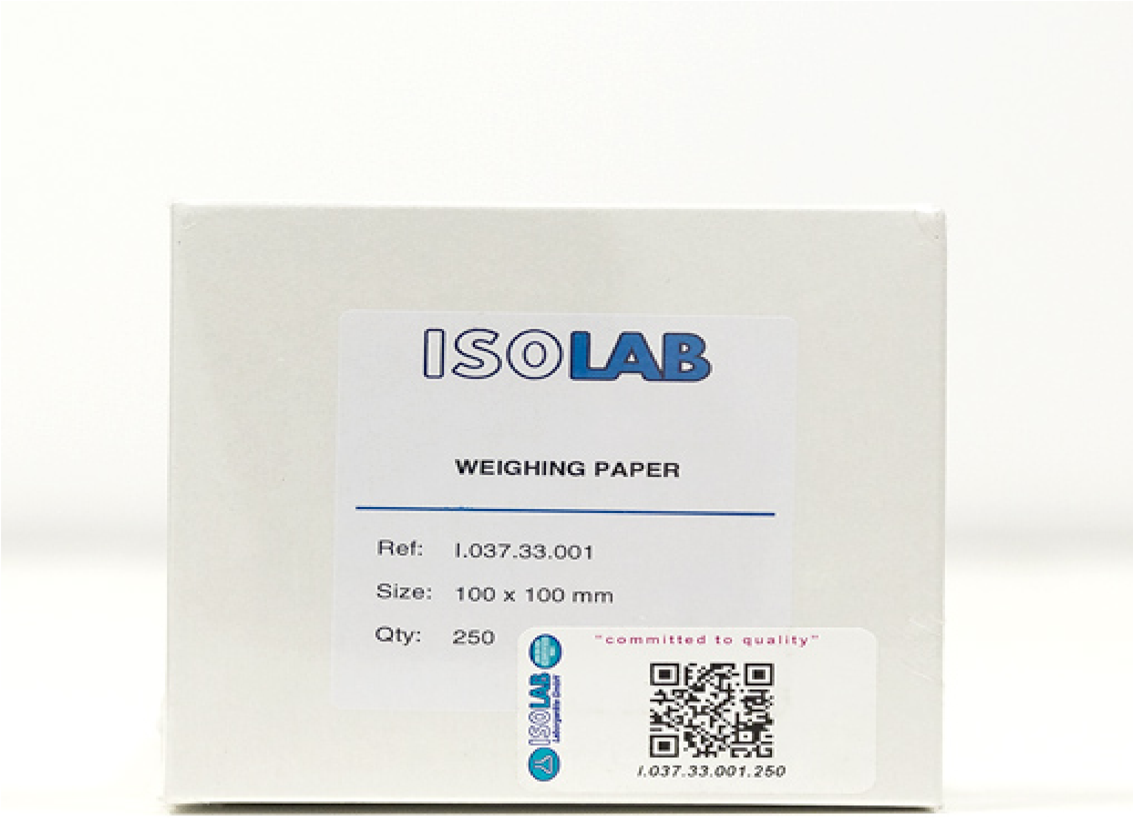

Figure 3.1 ISOLAB weighing paper [74].

We used ISOLAB weighing paper, which has very good properties such as high capacitance change when it detects moisture. We also tested seven different paper types to confirm that ISOLAB weighing paper is the best paper for the sensor. We produced multiple sensors with the same paper type to ensure the reproducibility of the results that we obtained. We compared the collected data from all these sensors and decided to use ISOLAB weighing paper. Sensors produced with different papers can be seen in Figure 3.2. In addition to tests on paper type, it was investigated how the sensors produced from the same paper but with different finger lengths would respond. Finger 1.7 indicates that the length of all fingers are 1.7 cm long, and the regular stencil means that every fingers are 1.5 cm long, which is the length we used for the last version of the sensor.

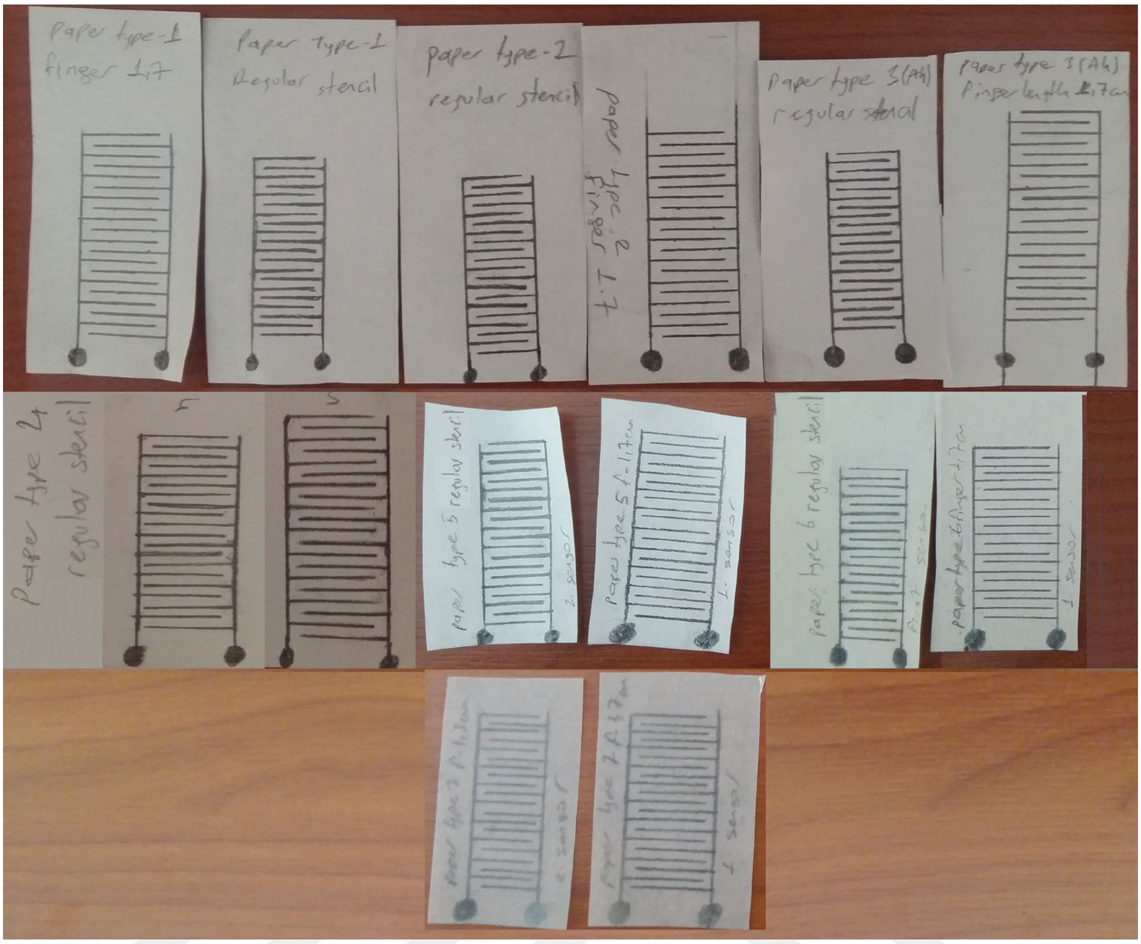

Figure 3.2 Produced sensors with different types of paper.

The obtained results with all these papers are given below. This data was collected when sensors were directly exposed to water vapor. We created water vapor by touching a soldering iron to a wet sponge. For respiration we would obtain a similar graph, the maximum peak points would correspond to exhalation, and the minimum peak points would correspond to inhalation. ISOLAB weighing paper corresponds to paper #5 in Figure 3.2. Some sensors produce high noise for various reasons and peaks cannot be seen clearly from the output graph. Other papers that can be used exhibit similar behavior to paper #5 such as having less noise, and peaks can be easily seen. Two different results are shown from produced sensors, except paper type 4, for the same paper and having the same finger lengths. It is aimed to show that the sensors are produced and tested in more than one, and similar results are obtained. The data here are preliminary studies showing the possible responses of sensors. Compared to the data here, much better results will be obtained after deciding on the paper and pencil for the sensor and the dimensions of the sensor. These results are given later in the thesis.

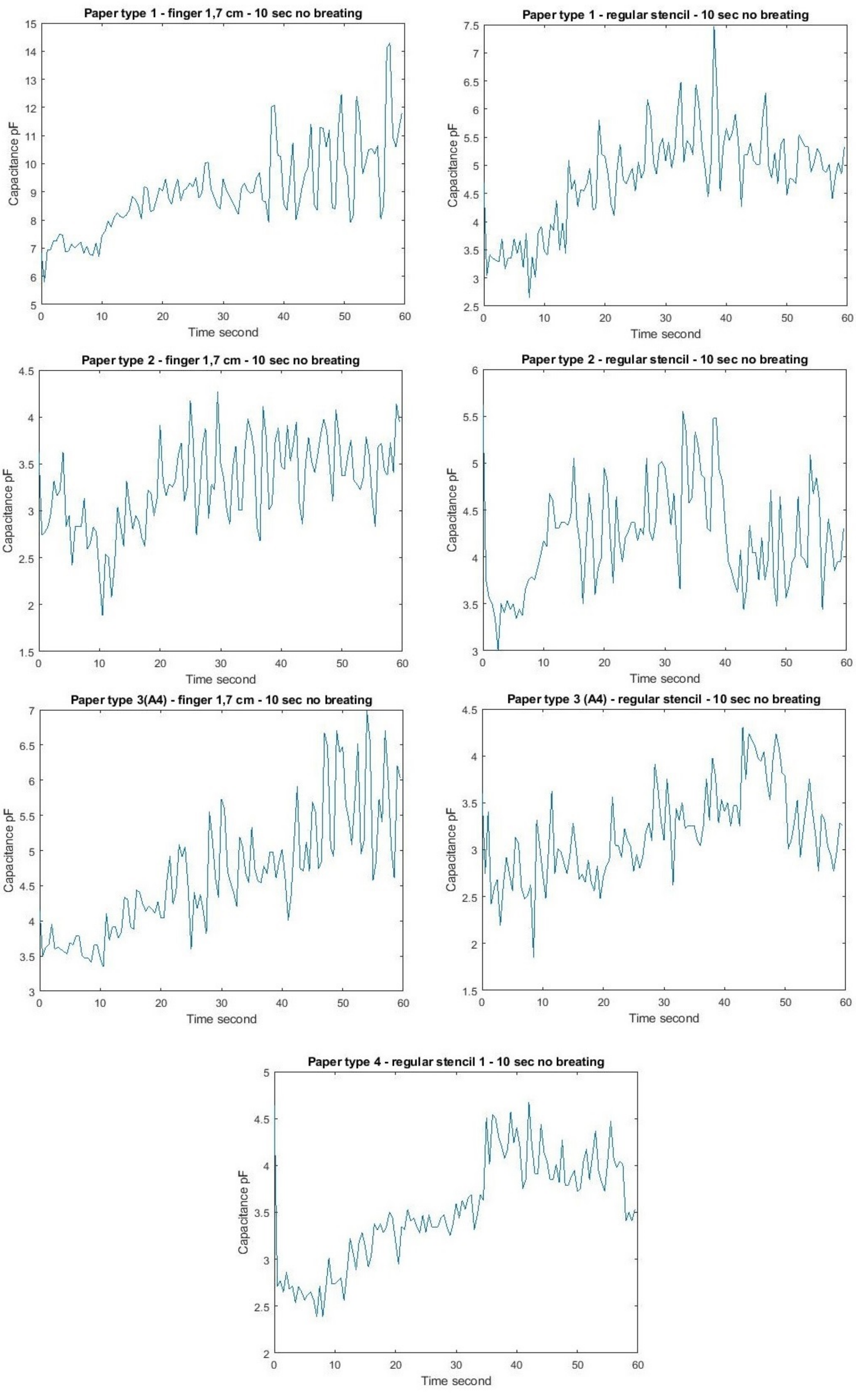

Figure 3.3 The results of sensors paper types 1 to 4

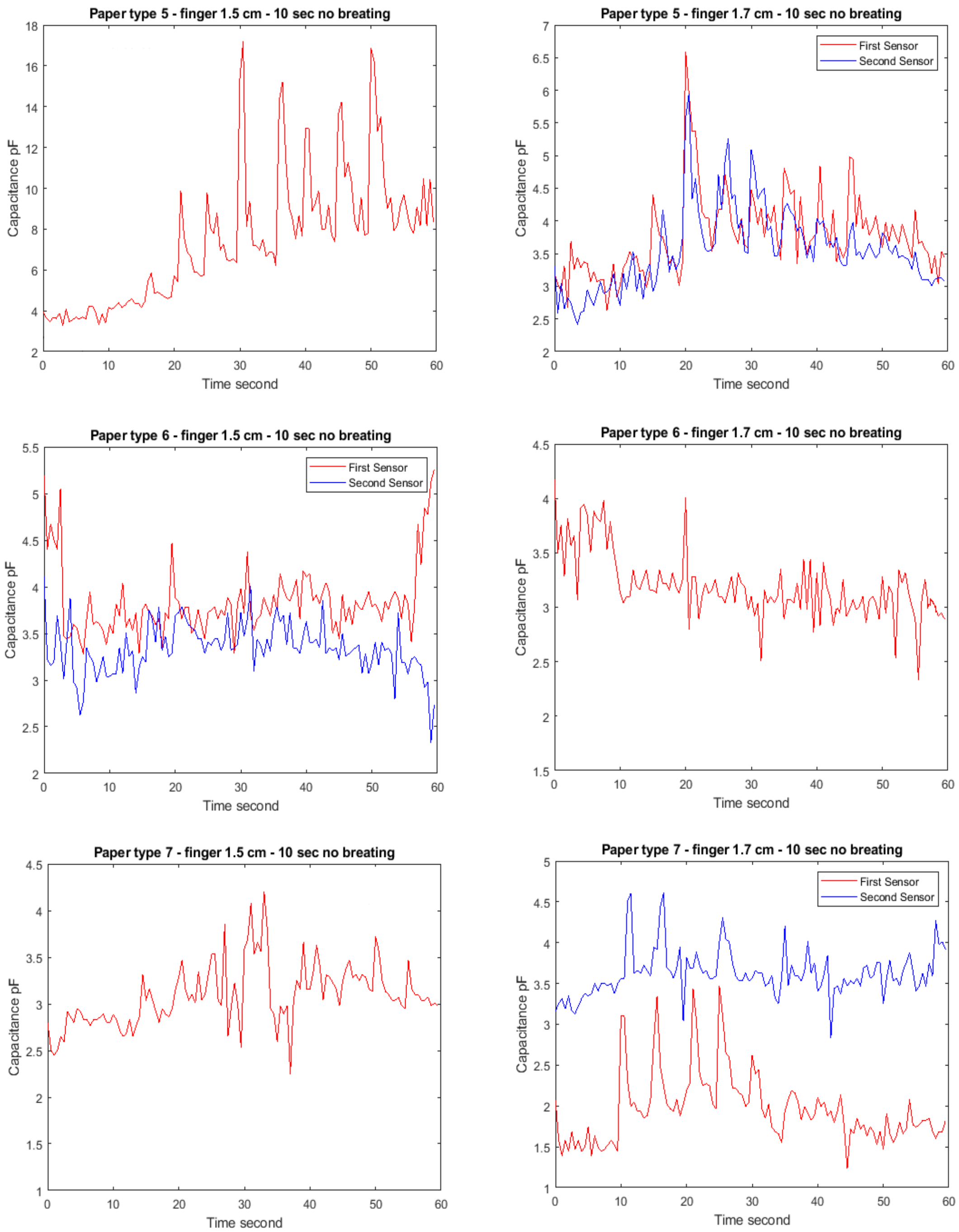

Figure 3.4 The results of sensors paper types 5 to 7

Paper type 1 is a paper that gives relatively better results. We can see peaks and capacitance change is higher compared to other papers. This paper is thin and moisture absorbent paper. Despite all these properties, it was not preferred because it is thicker and less sensitive than paper type 5. However, without paper type 5, it would probably be the main material for sensor production.

Even though paper type 2 is thinner than paper type 1, the results are not sufficient, as can be seen in the graph above, the capacitance change is not high enough since it cannot absorb enough moisture.

Paper type 3 is ordinary A4 paper. It is the second thickest and second least moisture absorbing paper among the other types, so it has one of the worst sensitivity. Therefore, it is not considered the main material for the sensor.

Paper type 4 is the thickest paper, and also it has the worst sensitivity. As can be seen from the figures given earlier, this paper is not suitable for sensor production due to the low capacitance change.

Paper type 5 is ISOLAB weighing paper, the paper currently used for sensor production. This paper is preferred because it is thin and makes a good moisture exchange with the external environment. It has high sensitivity and can make measurements at high speed. Moreover, the absence of saturation during respiratory monitoring, even in more extreme conditions which are given later, is a very important feature for the sensor. As can be seen Figure 3.4, the cleanest results belong to this paper type. Here, we can clearly see the response to the water vapor that we created by touching the soldering iron to the wet sponge. The maximum peak points show the moments exposed to water vapor. Considering all these, we decided to use this paper for sensor production.

Paper type 6 is a TUNA-SA branded paper sold under the Bowie dick Autoclave test pack name. It is yellow colored paper like straw paper. This paper was also not used because it was not sensitive enough and did not give the necessary response to the applied steam. It is one of the worst options for the sensor.

Paper type 7 is translucent paper. It's not thin enough and doesn't absorb moisture well enough either. More importantly, when we draw the pattern on it with a pencil, the paper does not hold the graffiti enough. We did not have such a problem with other papers. Therefore, it was difficult to make sensors with this paper. As can be seen from Figure 3, although paper type 7 is better than some paper types in terms of capacitance change, it was not preferred because paper type 5 performs better.

3.2 Pencils

Another important material used in the production of the sensor is the pencil [75]. We tried different pencils to obtain better results from the sensor and compared different kinds of pencils like mechanical pencils, pencils of various hardness, etc. Unlike the paper type, the pencil's effect on the sensor is limited. If the pencil is the kind of pencil that will leave a conductive material such as graphite on the paper, we can obtain a sensor that can detect moisture. Therefore, using different pencils didn't make much difference despite our best efforts. However, we tried to produce sensors using different kinds of pencils. Among all these different pencils, the most promising ones are Faber castell brand pencil with 2.5B hardness degree and Lyra brand pencil with 9B hardness degree. Although there is not much difference between them, we preferred the Faber castell brand pencil with a hardness of 2.5B. The results obtained with these two pencils are given in the graph below.

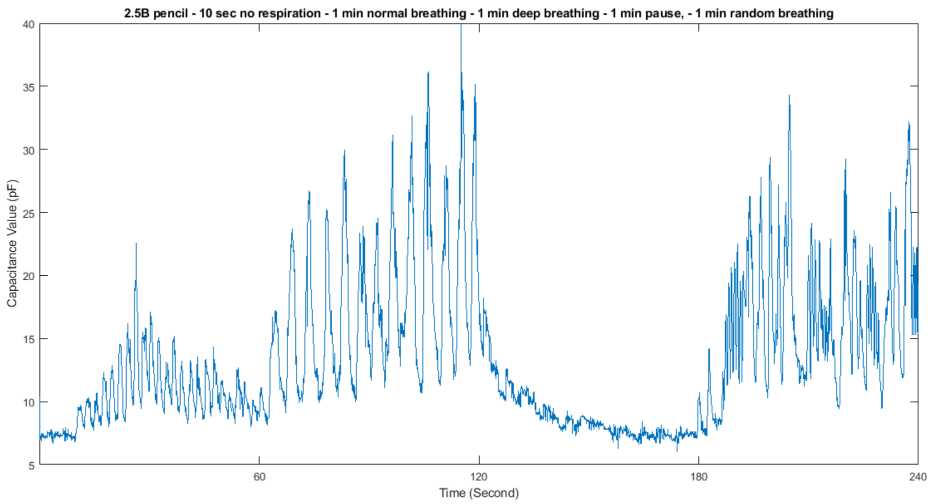

Figure 3.5 The measurement result of the sensor produced using a 2.5B pencil

The measurement method here differs from the previous soldering iron-sponge method. Here, the respiration data obtained from a real person are shared. The initial capacitance value of the sensor was measured for the first 10 seconds, then normal breathing was requested from the subject for up to 60 seconds. In the second 60 seconds, the subject started the deep breathing part. After 120 seconds, there was a 1-minute break, and the subject was asked to take random breaths in the last 1 minute. The Subject breathed only using his nose.

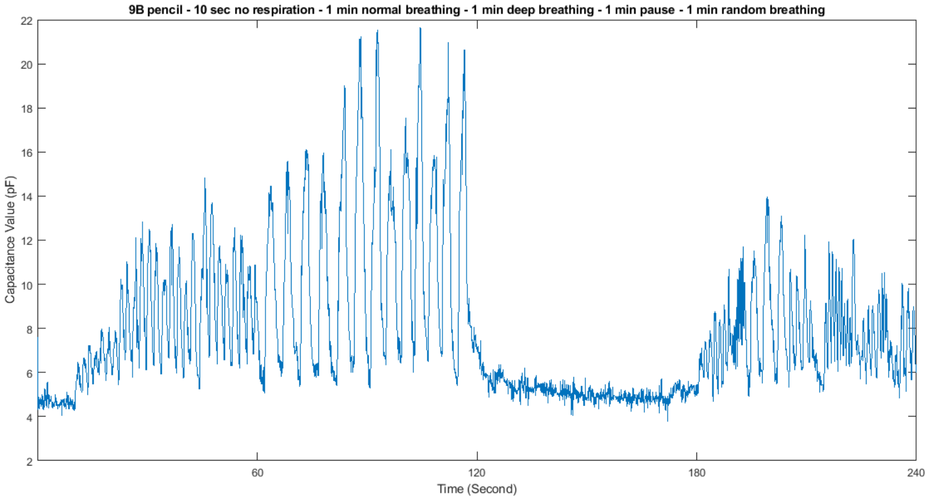

Figure 3.6 The measurement result of the sensor produced using a 9B pencil

Since the obtained results vary according to the measurement method, we tried to create the same environment as much as possible. We kept the distance the same and asked the user to breathe similarly for the two sensors. Despite our careful approach, there may still be errors, but due to the experiment being repeated several times, we decided that the 2.5B pencil is appropriate. However, sensors can be produced without any problems with both pencil types. Both produced sensors will be high quality sensors that can perform respiration monitoring.

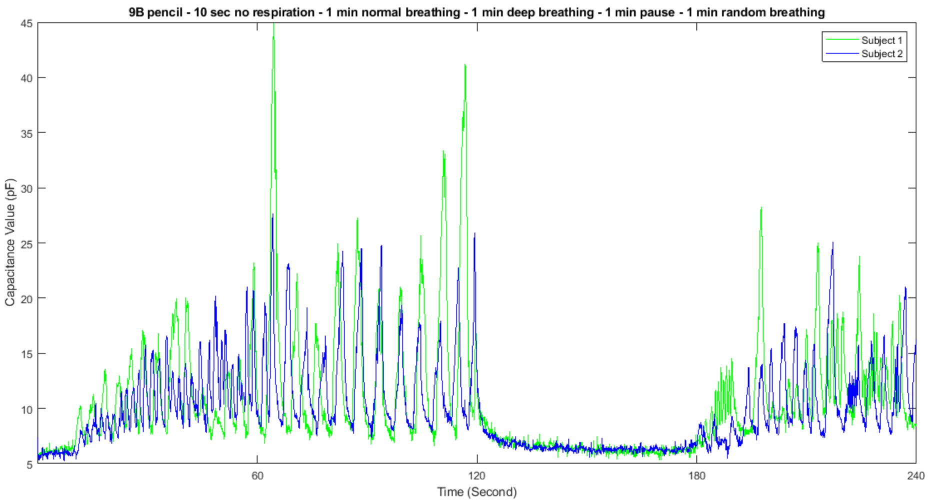

Figure 3.7 Results of different subjects using the sensor produced with a 9B pencil

Obtained data from different subjects using produced sensors with a 9B pencil can be seen in Figure 3.7. As can be understood from here, the sensor gives different responses to different people. We think that by using this feature of the sensor, it is possible to make comments about the health status of users.

3.3 Stencils

We used a stencil to ensure that the sensors are manufactured in a standard way. We placed this stencil on the sensor paper and drew the pattern on the paper. A transparent material called PMMA was used for making the stencil. For the sensor's final design, different sizes and different patterns are tested and the parameters were optimized. The optimum stencil was used to produce the moisture sensor. Considering factors such as sensitivity, production speed, and response time, we decided on the dimensions of the optimum stencil, and the sensor, as follows: finger lengths 1.5 cm, finger widths 0.1 cm, the total number of fingers 20, finger to finger air gap 0.1 cm, and finger to the lateral wall air gap 0.1 cm. The final form of the stencil can be seen in the figure below. In this subsection, the final stencil and sensor patterns, and their results are presented.

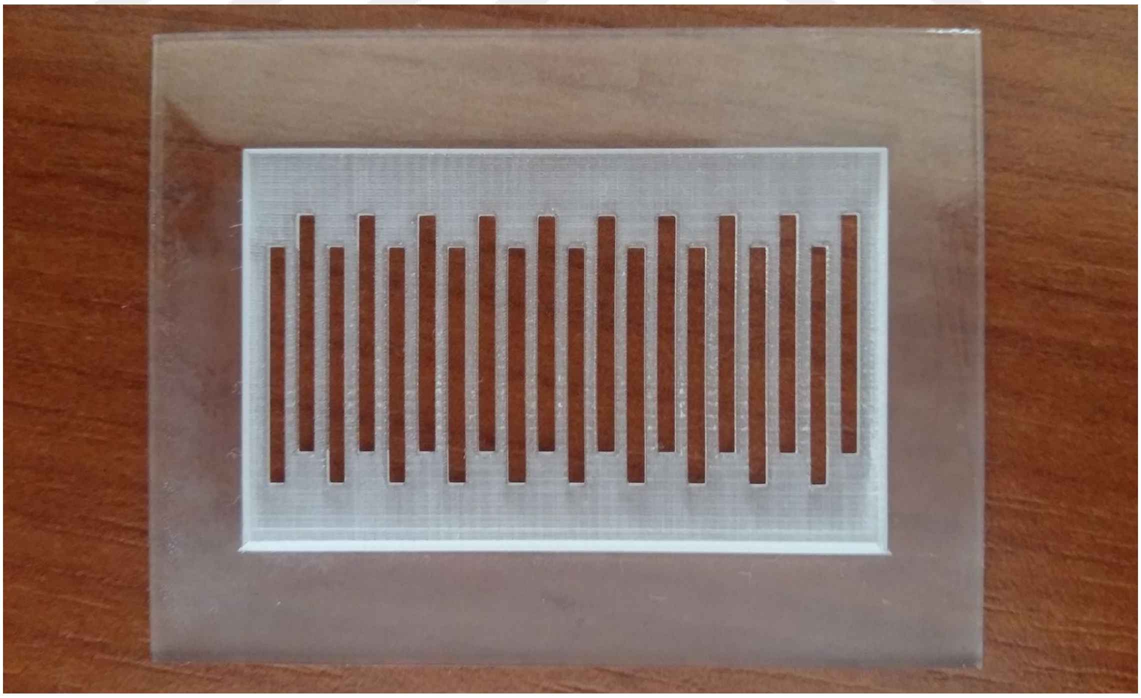

Figure 3.8 The final form of the stencil

3.3.1 Production of the Stencil

Epilog engraver zing laser printer was used for stencil production. Different patterns were designed with CorelDRAW X7 software and they were produced using 2 mm thick PMMA material. The laser printer has two different operating options. One of them is raster mode, and the other is vector mode. In vector mode, the material is completely cut to reveal the pattern, while in raster mode, a part of the material is burned, reducing its thickness as desired. For example, while the PMMA we use is 2 mm, we can reduce this thickness to 1 mm with raster mode operation. We have reduced the thickness of the PMMA by using the raster option so that the pencil can fit into the stencil's gaps more easily. In Figure 3.8, the shaded area in the middle that appears different is the area with reduced thickness. In order to make PMMA stencil more durable, we did not apply the raster process to the area outside the shaded area. Thus, we could draw the pattern more easily with a pencil, and PMMA would not break easily. Using the vector option, the laser printer cut the PMMA in the form of the designed pattern. Before coming to the final design, many practical and unpractical designs were made. One of them is given in the following figure.

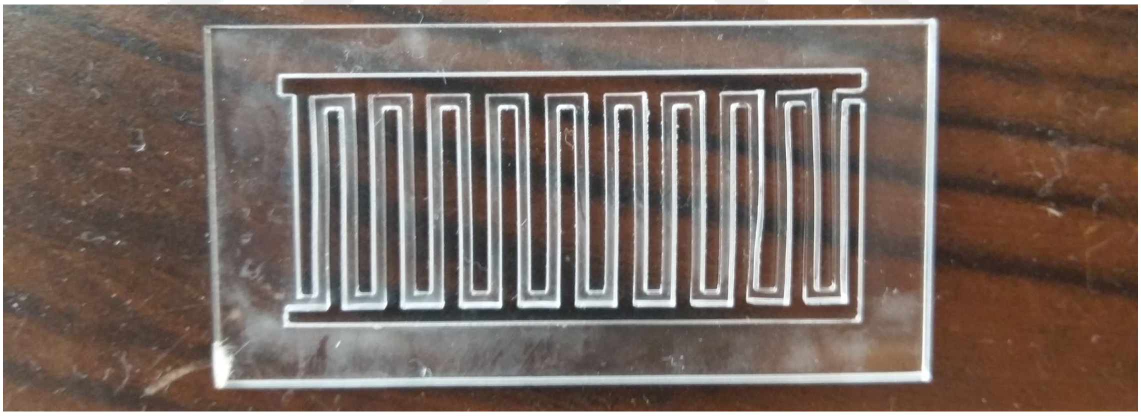

Figure 3.9 One of the stencil designs

This design has extremely flexible fingers, as can be seen from the figure. Although better results could be obtained by using different materials, we continued to use PMMA, instead, we changed the design and obtained a more robust stencil.

3.3.2 Finger lengths

We have shared the size of the stencil before. To mention it again, they are determined as follows: finger length 1.5 cm, the total number of fingers 20, finger to finger air gap 0.1 cm, and finger to the lateral wall air gap 0.1 cm. The sensor has a total length of 5 cm and a total width of 3 cm. The dimensions with the best sensitivity of the sensor are given above, we made sure of this by making many experiments with different dimensions beforehand. The ISOLAB paper was named paper type 5. The tested different finger sizes using paper type 5 are given in Figure 3.10.

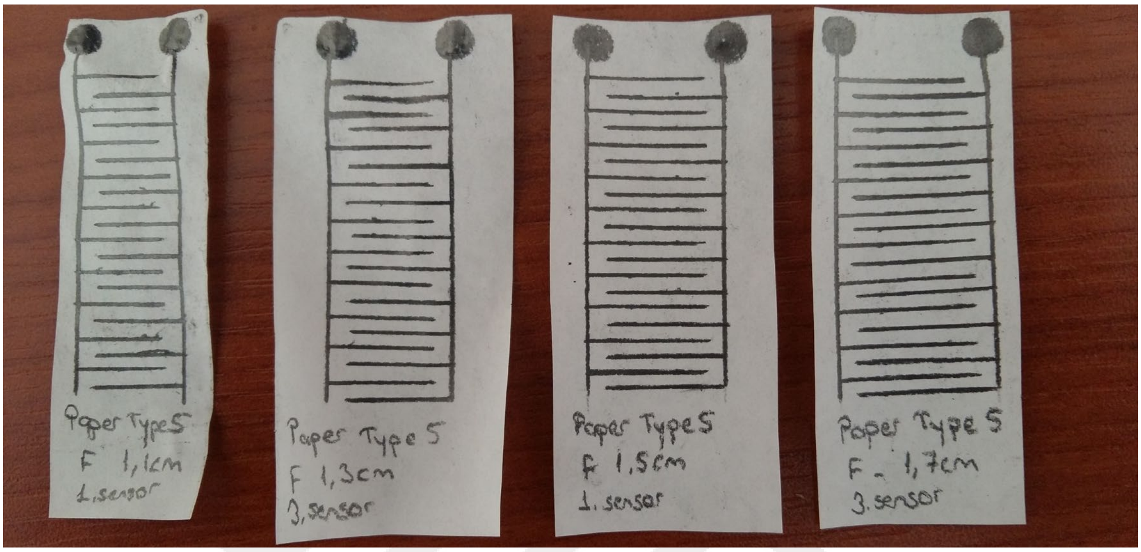

Figure 3.10 Sensors with different finger lengths

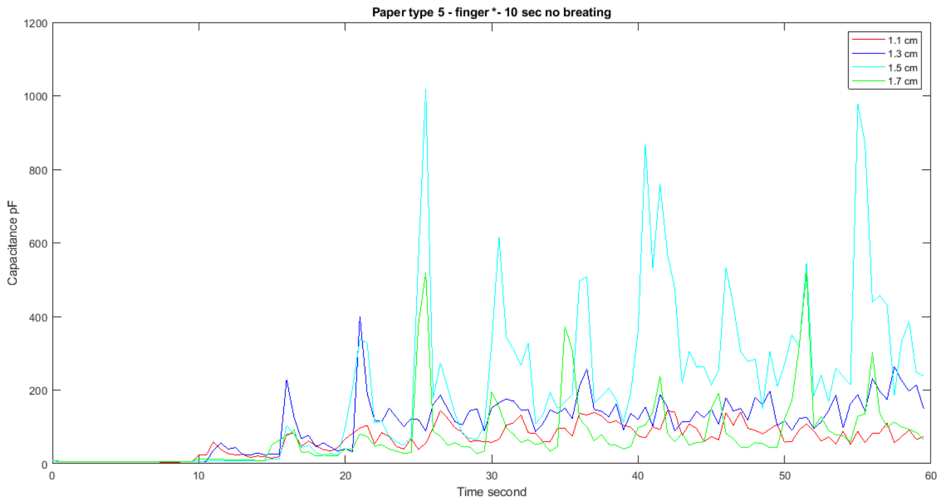

Figure 3.11 Experimental results of sensors with different finger lengths

In Figure 3.10, there are sensor numbers like 1. sensor or 3. sensor. This shows the produced sensor number that is 1. sensor is the first produced sensor for finger length 1.1 cm similarly 3. sensor is the third sensor produced for the finger length 1.3 cm. The measurements were repeated several times to make sure that sensors have high repeatability and reproducibility. The capacitance changes according to finger length can be seen in Figure 3.11. It is obvious that the sensor with a finger length of 1.5 cm has a definite advantage over other sensors with different finger lengths. While the sensor with a finger length of 1.5 cm had a capacitance value that increased up to 1000 pF, the sensor closest to it could only reach half of this value. In order to obtain these values, the initial value of the sensor was observed for the first 10 seconds, and then the process was started. Although the first 10 seconds seem as if no value was observed, in fact, values around 5 pF were measured in the first 10 seconds. The reason why it is seen as 0 in the first 10 seconds is due to the scale of the graph. Values that fluctuate around 5 pF seem to be a constant 0 pF because the difference between the baseline and the peak point is very large. For the measurement, water vapor was applied to the sensors once in 5 seconds and the response of the sensors was observed. Sensors respond very quickly due to successful paper selection.

3.3.3 Pattern of the sensor

We used interdigitated fingers pattern for the sensor. In the literature, this pattern is generally preferred more than other pattern types. The interdigitated pattern is more advantageous than other types, because it has higher sensitivity compared to other patterns and low noise in the signal produced by the sensor. However, we tested different patterns to get the best performance from the sensor. We researched how the sensor behaves against the different patterns we designed. We were able to detect respiration with all the patterns we tried, but their sensitivities varied considerably, depending on the pattern. In some patterns, the sensitivity was extremely low. We preferred the noise-free pattern with the highest sensitivity. For the measurement, the initial value of the sensor was observed for the first 10 seconds, then a normal breath was taken for up to 1 minute, followed by deep breaths. After waiting for 1-minute, random breathing was taken and the measurement was completed. The results of our research on interdigitated, spiral, and eye pattern sensors can be found in this section. The pattern with an interdigitated design and the obtained result are given in the figures below.

Figure 3.12 Interdigitated fingers design

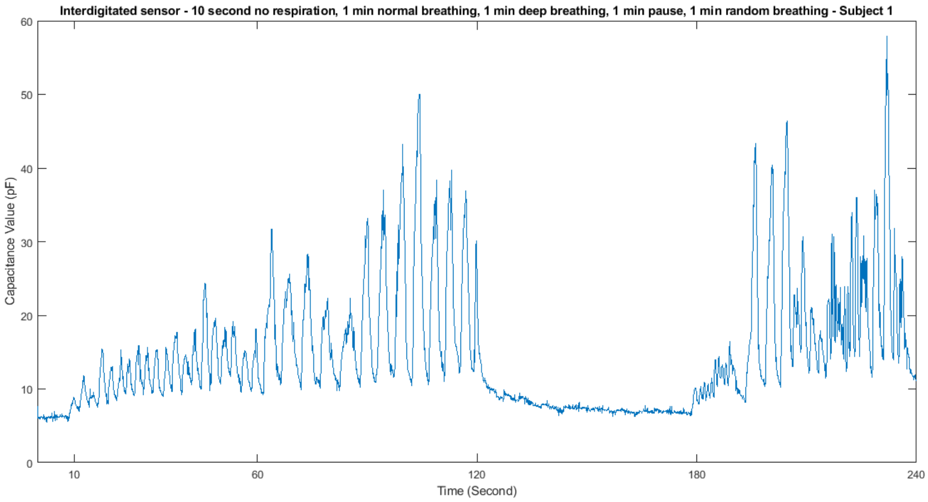

Figure 3.13 The Experimental result of interdigitated sensor

The spiral pattern has less sensitivity, and it has a noisy signal than the interdigitated pattern. Although the spiral sensor can detect respiration, it is obvious that the interdigitated pattern is better. The experiment method is the same as for the interdigitated sensor. Figures of the sensor produced with this pattern and the result of the experiment can be seen below. In order to better understand the difference, the graphs of the interdigitated and spiral sensors are given separately. The interdigitated sensor went up to 50 pF, while the spiral sensor went up to 11 pF. This sensor, which is also relatively difficult to manufacture, can be used in respiration detection after various improvements.

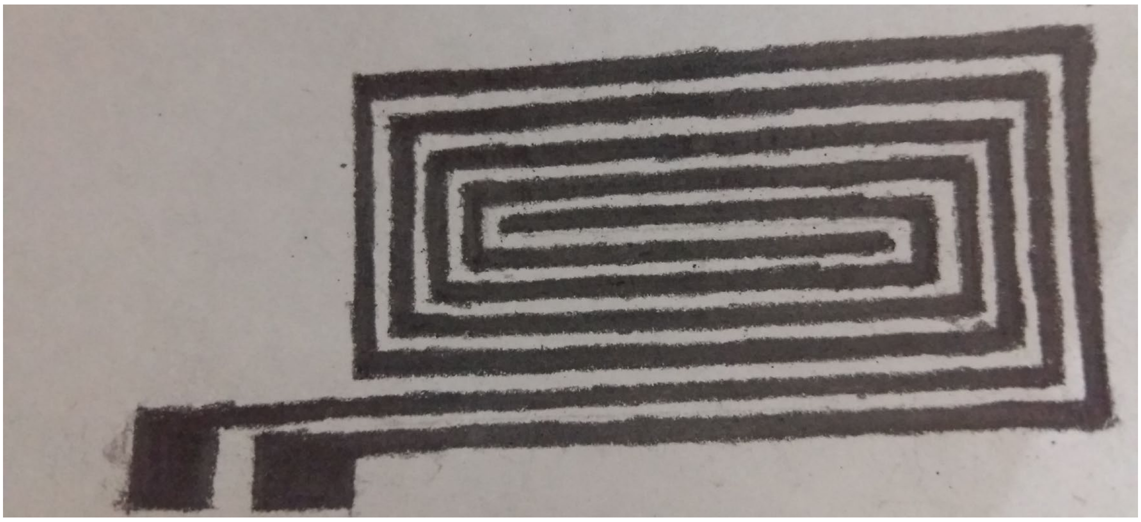

Figure 3.14 The spiral sensor

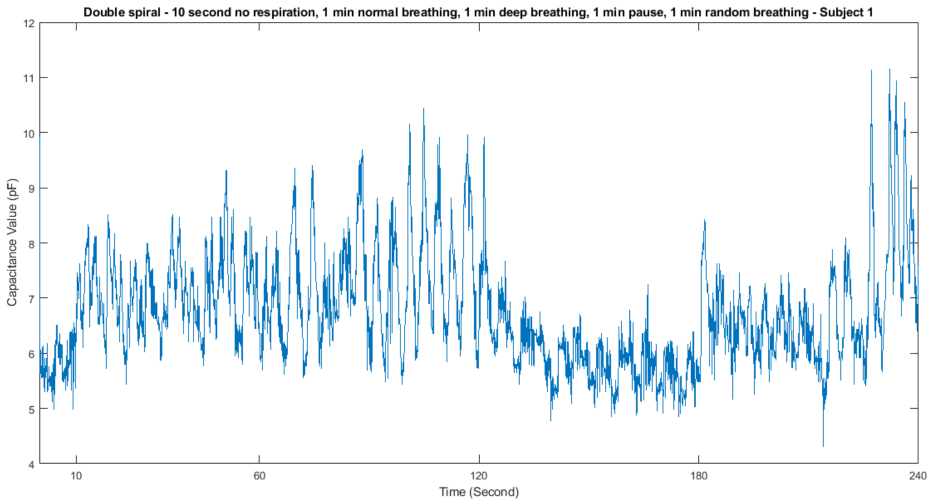

Figure 3.15 The experimental result of the spiral sensor

The last pattern to be mentioned in this subtitle is the eye pattern. This pattern has the lowest sensitivity and it has the highest noise among the 3 shared patterns. While it can detect humidity, it's a pretty poor design for respiration monitoring. Therefore, for this design, the sensor was investigated with a method other than the method used for the two above. The sensor was exposed to water vapor every 5 seconds with the soldering iron sponge method, which was used before. The sensor and the obtained results from this experiment can be seen in the figures below. Finally, ISOLAB weighing paper is used in all designs mentioned in this section.

Figure 3.16 The eye sensor

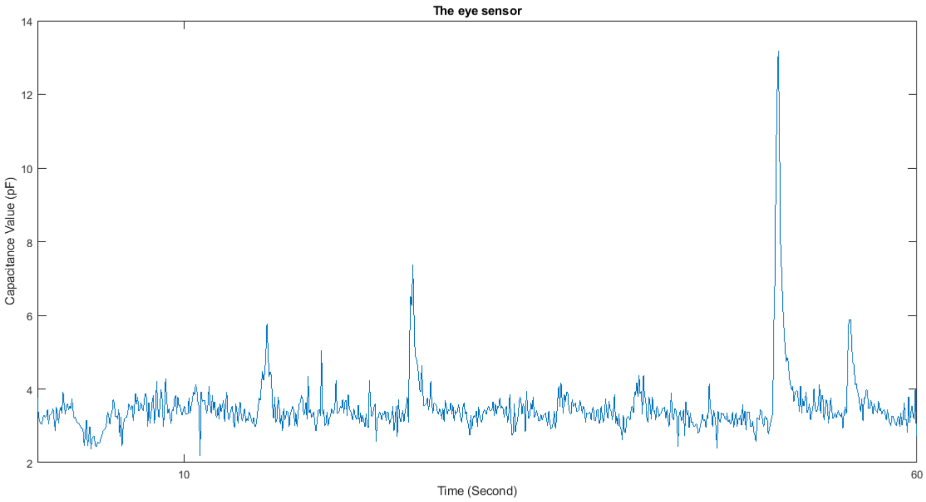

Figure 3.17 The Experimental result of the eye sensor

3.4 Sensors

The sensor has a total length of 5 cm and a total width of 2.1 cm. The sensor responds differently to the breath coming out of the mouth and to the breath coming out of the nose. Since the moisture content in the mouth breath is usually higher, measurements will be made using only the breath coming out of the mouth. Also, the result obtained from the sensor varies according to the distance of the sensor to the mouth. We used a mechanical design produced in a 3D printer to keep the distance constant. The details of the design can be found in Section 3.6 Mechanical Parts. For this design, we extended the width of the sensor a little more than it should be to place the sensor inside a mechanical part called the sensor holder. So, we have an extra 0.9 cm of space to grasp the sensor. The last version of the sensor can be seen in Figure 3.18.

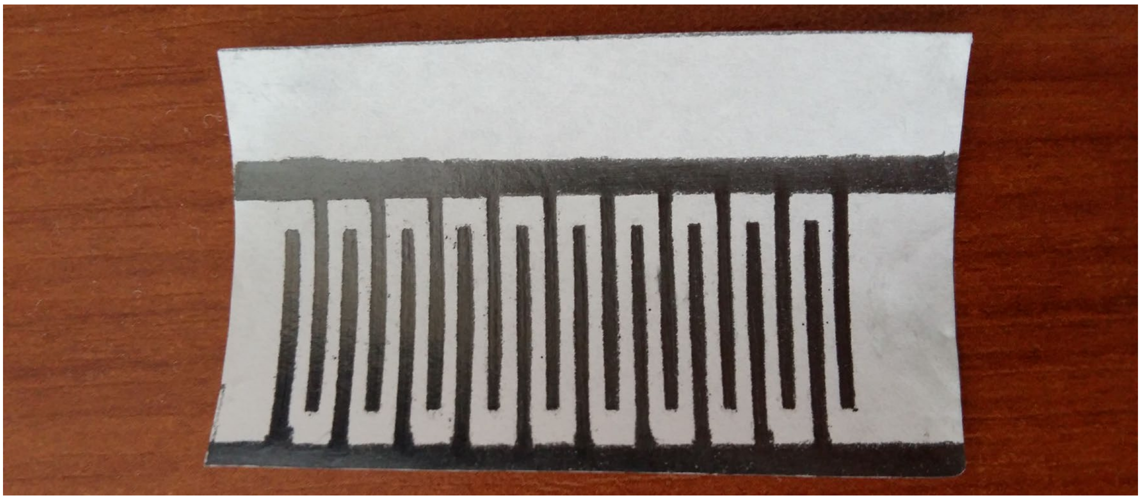

Figure 3.18 The last version of the paper-based moisture sensor

There was no pattern on the back of the sensors we designed, so the paper remained blank as it was. In the literature, there are sensors painted back side [47]. We tried this method to improve the sensor. After this process, we expected a sensor that has a higher capacitance change, and the result was not exactly what we expected. The capacitance value, whose peak point was around 25-30 pF before, increased up to 55 pF with this method. However, we obtained pulse-like noise signals from the sensor we produced with this method. Signals obtained from the sensor had an oscillation similar to a square wave. In Figure 3.19, these signals obtained from the sensor can be seen along with the front and back sides of the mentioned sensor.

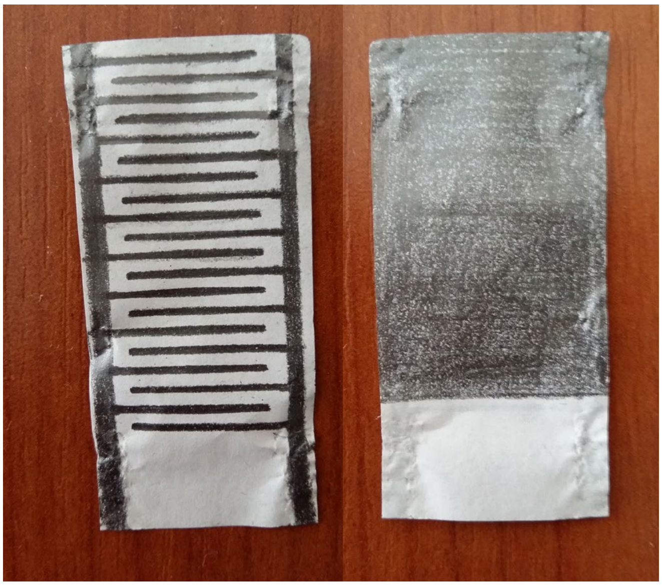

Figure 3.19 The back side painted sensor A) Front side B) Backside

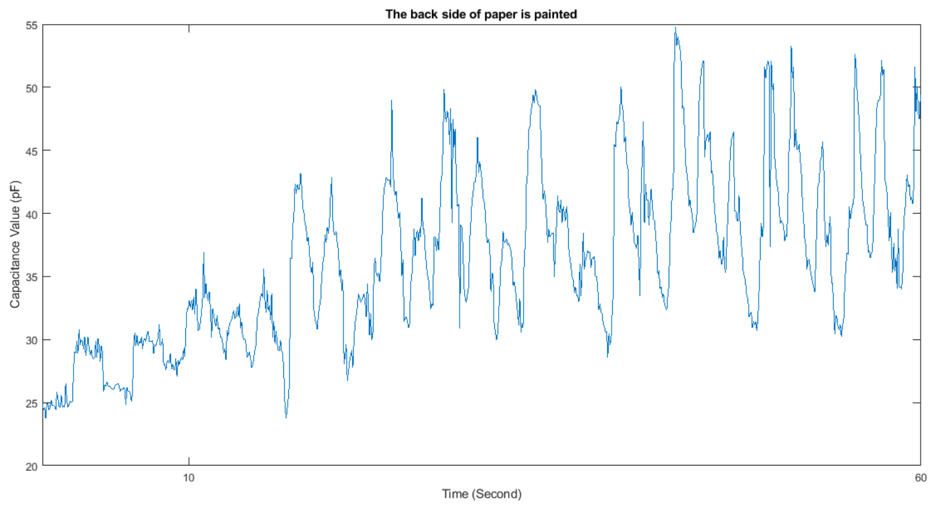

Figure 3.20 The signal obtained from the back side painted sensor

We stopped painting the back to produce a noise-free sensor. Instead, we drew the same pattern on the back side of the paper. In this drawing, due to the geometry, the air gaps between the two fingers on the back side and the front side of the sensor coincided exactly, but the air gaps between the side walls and the fingers covered each other. We connected Arduino to the pattern on the back and the pattern on the front at the same time using crocodile cables. Although the improvement in the sensor is much better, the square wave showed itself in this design as well. The mentioned design and the result obtained from this sensor are as in the figures below.

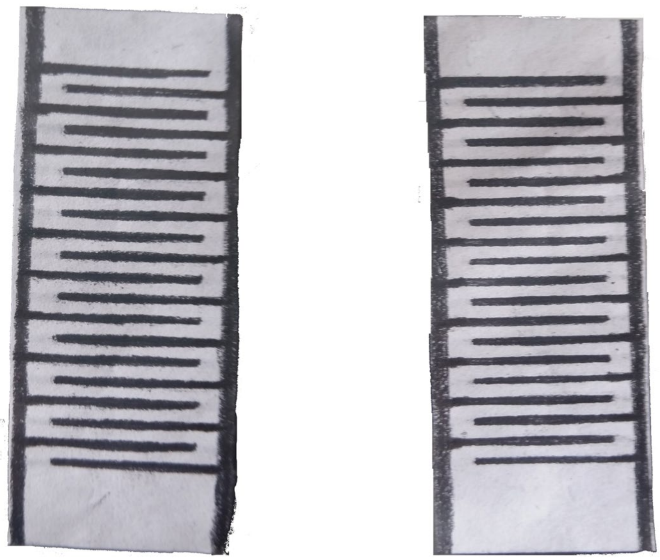

Figure 3.21 The sensor with the same pattern on both sides A) Front side B) Backside

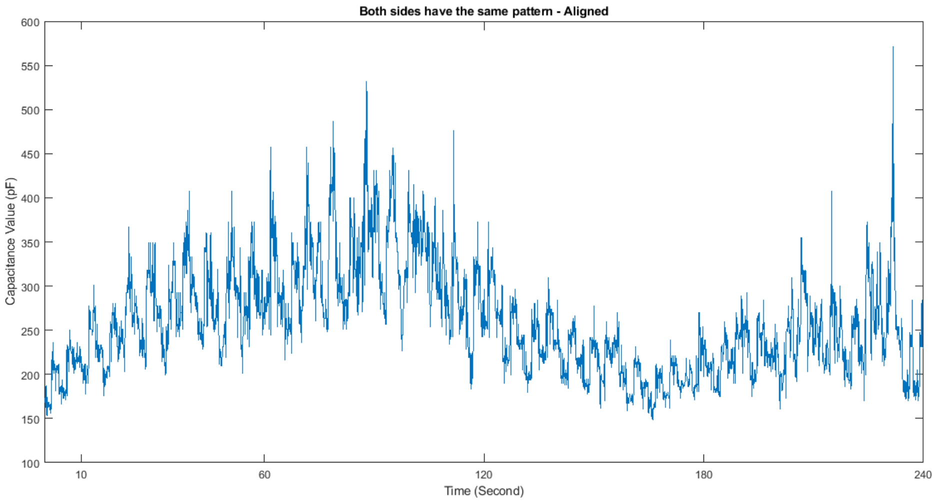

Figure 3.22 Experiment result of the sensor with the same pattern on both sides

After giving up on this design too, we drew the normal pattern on the front of the paper and the mirror image of the pattern on the back side of the paper. Therefore, the used stencils are mirror images of each other. Thus, both the air gaps between the back side fingers and the front side fingers and the air gaps between the fingers and the side walls coincided. We tested the sensor after completing the connections, again using a crocodile cable. In the end, we achieved improvements with this design and got rid of the square wave, but due to the difficulties in the production of this sensor, we returned to the design with a blank side and a pattern on the other side. The mentioned design and the obtained result from this design are as follows.

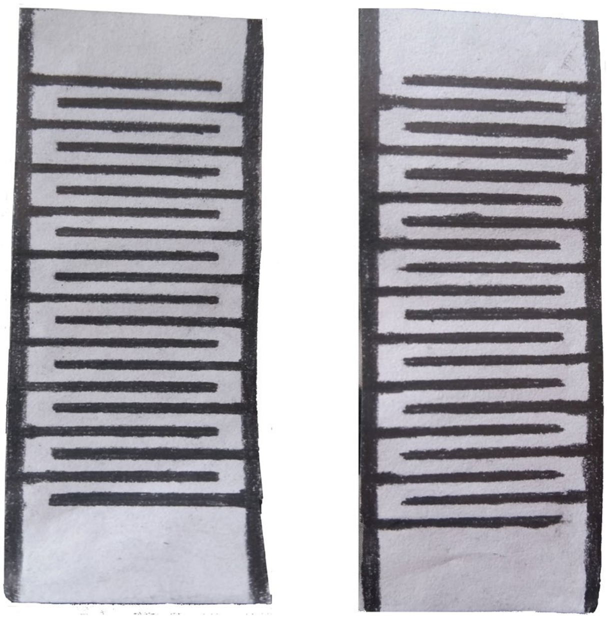

Figure 3.23 Normal and mirror image patterned sensor A) Front side. B) Backside

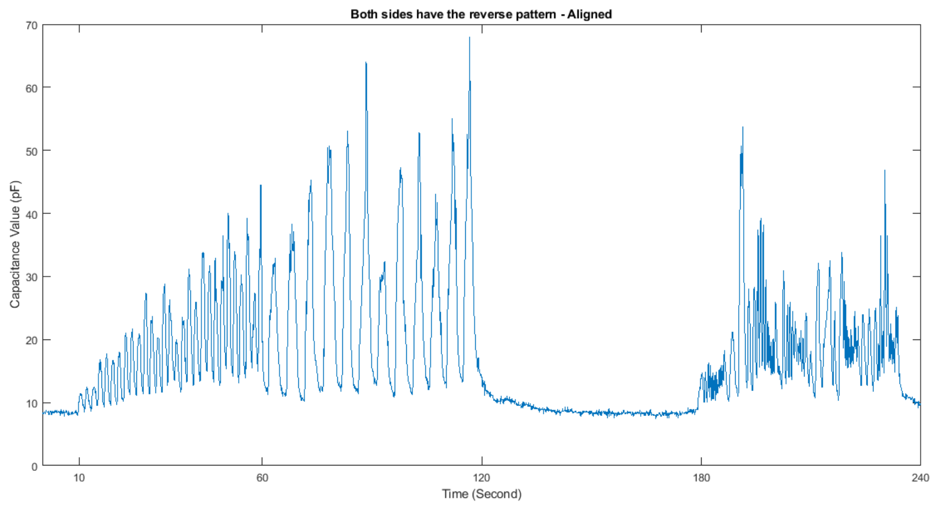

Figure 3.24 Experiment result of the sensor with reverse patterned

Although the sensor is designed as disposable it can be used multiple times and for long periods. To test the sensor, it was used for at least 12.5 minutes at different times during the day, and then the sensor was left for one day. In the end, the sensor preserved its functionality both during the day and after the waiting period. This process has been performed several times. It has been used at least once a week for 3 months and has been kept in such a way that the sensor will not be damaged. Consequently, the sensor continued to detect respiration. At the end of 3 months, obtained data from the retested sensor is as follows.

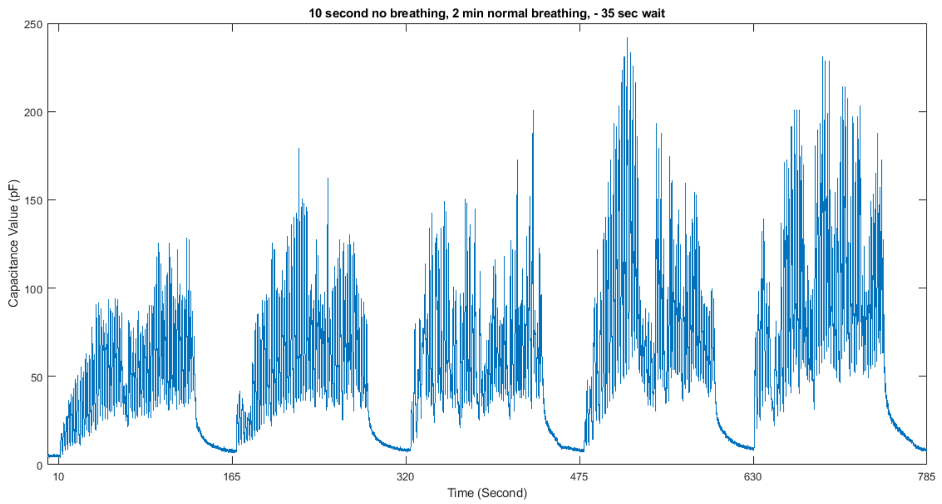

Figure 3.25 The experimental result of the sensor at the end of the 3-month period

It is clearly seen here that the sensor can maintain its function for long periods of times as long as there is no physical damage. As mentioned above, although the sensor is for single use, we carried out this experiment to understand the sensor's characteristics. We tried to measure the lifetime of the sensor and from the result, we found that the sensor can be used for more than 3 months.

We wanted to compare the results that we obtained from the paper-based moisture sensor with a different commercial humidity sensor, and we decided to use the DHT22 sensor which is easily available through online market. DHT22 is a sensor that can measure both humidity and temperature. It can measure the temperature with a very small margin of error, +/- 0.5°C. For humidity measurement, although the error margin increases and measures with +/-5% accuracy [76], it can be considered an acceptable value for the measurement. We prepared an experimental setup to compare the DTH22 sensor and the sensor we produced. For this experiment, we positioned the DTH22 sensor and the moisture sensor in a closed box with small ventilation holes on it in such a way that they stand side by side with each other. Thus, the amount of moisture detected by both sensors would be approximately the same. We sent the water vapor obtained with the sponge soldering iron method into the box every 5 seconds. In Figure 3.26, the experimental results and comparison of the DTH22 with the moisture sensor are given.

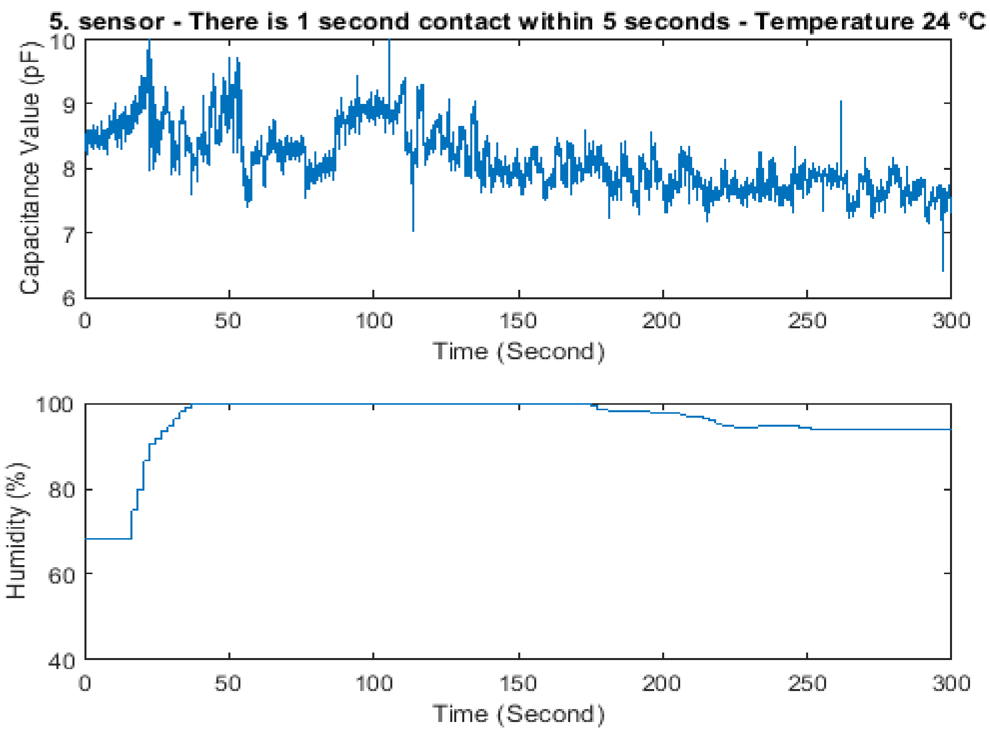

Figure 3.26 The experimental results and comparison of two sensors. The upper one is the sensor that we produced below one is the DTH22 sensor

When exposed to water vapor, the DHT22 sensor was saturated immediately and remains saturated for a long time. The advantage of our sensor is clearly visible here. While the moisture sensor detects the water vapor and it is ready to measure again, DHT22 becomes saturated for a long time and cannot measure the changes. Even if we do not expose DHT22 to water vapor after being saturated during the experiment, it remains saturated for a while. One conclusion we can draw from this experiment is that our sensor is a sensor that works fast and remain unsaturated under extreme conditions.

Another factor affecting the sensitivity of the sensor is the width of air gaps. Air gaps in the final version of the sensor have a width of 1 mm. When we kept the sensor dimensions constant, which is 5x2.1 cm, and produce the sensor with an air gap of 2 mm, the sensitivity of the sensor decreased. Even if the decrease in the number of fingers has a negative effect on this result, we were able to make such a comparison since we did not change the length and width of the sensor. If we kept the number of fingers constant, the size of the sensor would increase a lot, so we decided to keep the size constant. We thought that we can achieve better results if we reduce the length of the air gaps, but since the dimensions are very small, it would be difficult to produce, so we decided to continue with a sensor with 1 mm air gap widths. Consequently, we can say that the sensitivity decreases gradually as the air gap width increases for the same sensor sizes. The 2 mm air gap sensor shown in the figure below has 12 fingers.

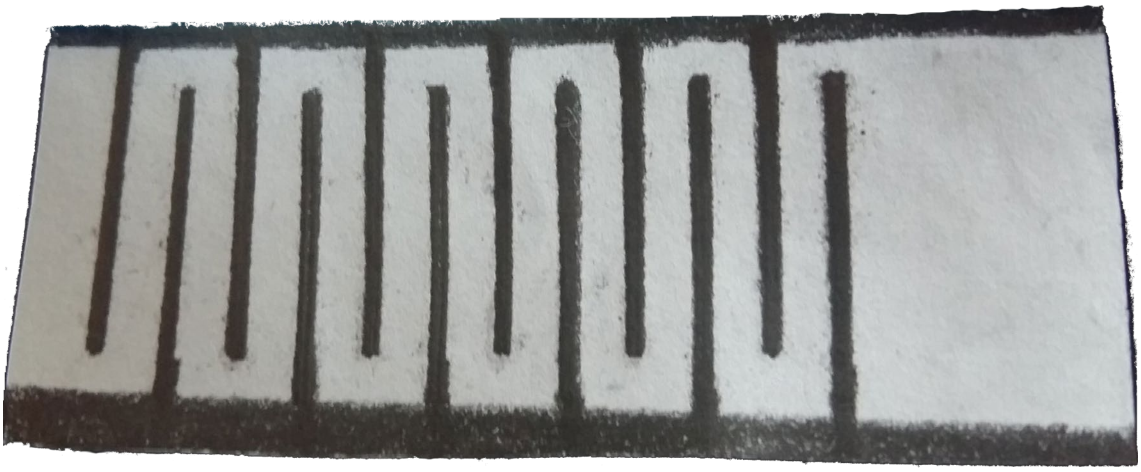

Figure 3.27 The sensor with 2 mm air gaps

The experimental results of the sensor with 1 mm air gaps and the sensor with 2 mm air gaps are given in the same figure below. As can be clearly seen from here, 1 mm air gaps give us better results. As mentioned above, although smaller sized air gaps will probably give us better results, we have decided on the final air gap length of the sensor which is 1 mm since it is relatively easy to manufacture.

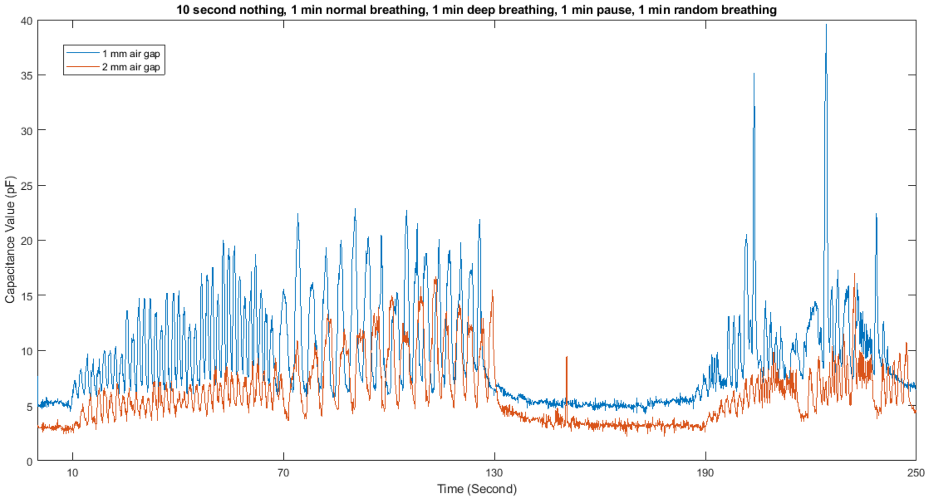

Figure 3.28 The experimental result of the sensor 1 mm air gap vs. 2 mm air gap

We mentioned that the sensor gives different responses to different people. This is due to the difference in moisture content in the breath of different people. The experimental results of two different people are given in Figure 3.29. In the figure, subject 1 generally achieved higher capacitance values than subject 2. This means that the breath of subject 1 has higher moisture content. Since the sensor is disposable, the experiment was conducted with two different sensors. The initial values of the sensors are very close to each other and are around 5 pF. This value fluctuates by a very small amount for various reasons.

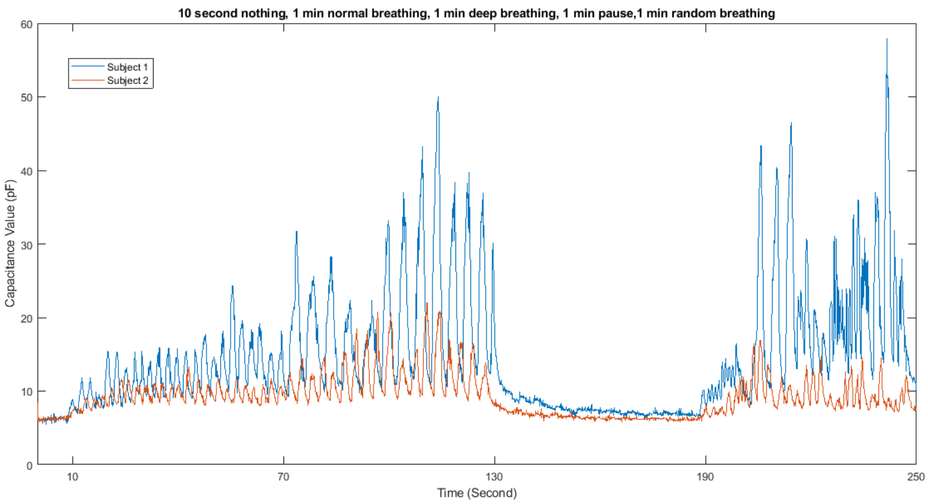

Figure 3.29 Experimental results of subject 1 and subject 2

We obtained different results even from the same person according to the use of mouth or nose. For the same person, the difference in the moisture content from the mouth and nose causes such a result. According to observation from the experiments, we generally obtained higher capacitance values when the mouth breath test was performed compared to the nose breath test. Since the difference is too much, we can give the results in different figures. Otherwise, the obtained result using the nose would look like a straight line due to the difference. The results are as follows.

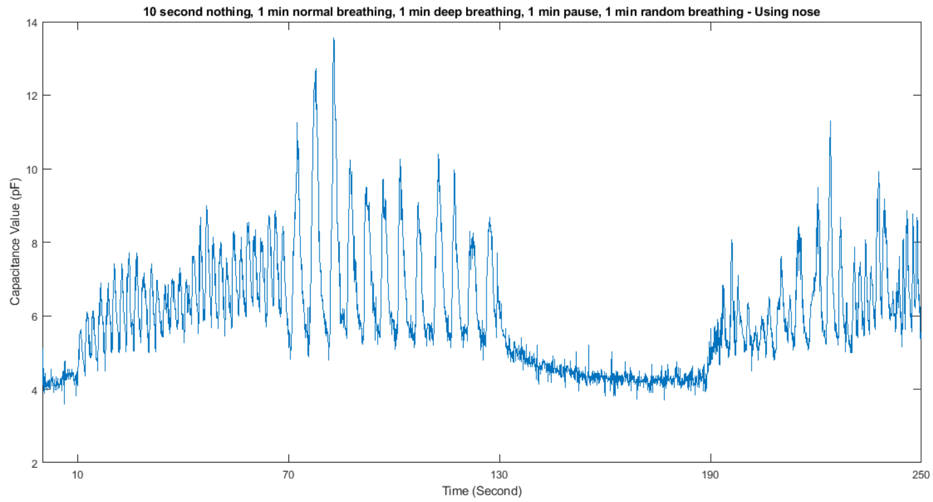

Figure 3.30 The result obtained by breathing through the nose only

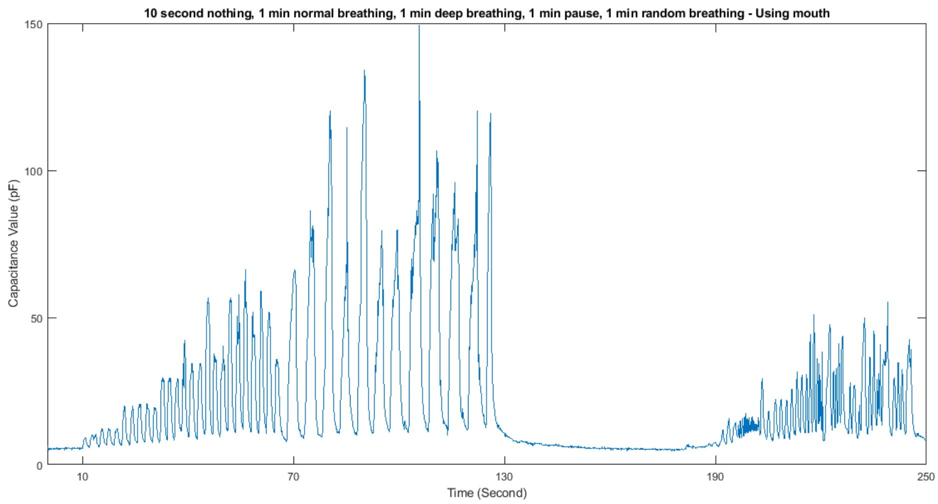

Figure 3.31 The result obtained by breathing through the mouth only

3.5 Electronic Circuits

We needed a microcontroller to process the data from the moisture sensor. At the beginning, we decided to use the Arduino Uno, one of the popular microcontroller boards. For capacitance measurement, we used the codes referred to as capacitance meter on the Arduino website [77]. With these codes, Arduino Uno detects the capacitance value by calculating the time constant. The time constant is expressed by the following formula.

$$\tau = RC \qquad (1)$$

Once Arduino determines the time constant, it can calculate the capacitance value very easily. For relatively high values such as microfarad and above, it is necessary to add a resistor to the circuit, but for smaller values such as picofarad, measurements can be made without adding a resistor. Since our sensor reaches values around picofarad, there is no need to add an external resistor, in this way the circuit remains quite simple and easy to set up. All we need to do is to connect the A0 pin of the Arduino to one terminal of the sensor and the A2 pin to the other terminal of the sensor so that we can measure the capacitance value of the sensor. To ensure the accuracy of the measurement, we made measurements with capacitors whose values are known. The results were quite close to the known values of the capacitors. After realizing that the small differences are due to the error margins of the capacitors, we used Arduino Uno and ready-made codes for a while for our work. When we decided to collect data from different people, we printed a 3D design using a 3D printer to adjust the distance between the user's mouth and the sensor. The information about this design is provided in the next section. Since Arduino Uno will take up a lot of space in this design and we want to use the design as wearable, we decided to use Arduino Nano instead of Arduino Uno so that we could do the same job with much less space. Then we modified the codes according to our needs. To compare the sensor, we produced with the DHT22 sensor, we wrote code to interpret the data from both sensors. Then we used Arduino Nano’s built-in LED which is connected to the 13th pin of the Arduino, in order to understand that the sensor and the Arduino have electrical contact. By using this method, if the sensor is connected to the Arduino, the LED will turn on, otherwise, it will turn off. We fixed the Arduino to the printed part on the 3D printer and completed the electronic design. The final version of the code we used for the capacitance measurement is given in the appendix.

3.6 Mechanical Parts

We needed mechanical parts to fix the distance to standardize the measurements. Thus, the distance from the mouth to the sensor will be constant for different subjects, and we will take measurements very easily with a wearable system. We designed a head mask/holder and printed it using 3D printers. The subject wears the head mask similar to face shields. The system is designed to start taking measurements as soon as it is worn. Therefore, we have obtained a system that is very easy to use. The design consists of two parts, the lower part, and the upper part. The lower part can hold the sensor stably in its part called the sensor holder. In addition, the sensor can be electrically connected since it requires electrical contact with Arduino Nano through cables. Apart from this, the lower part can move up and down so that the sensor position can be adjusted to the mouth level. The upper part has a mechanism that enables the movement of the lower part. Thus, the position of the sensor is adjustable up and down. Arduino Nano is also located on the upper part. After adjusting the up and down position, we will be able to get reliable results from users. Finally, it has a rubber band that can be adjusted according to the size of the heads of different users. Thus, the entire system is wearable. An image of the design and usage example are given in the figure below.

Figure 3.32 A) The head mask for the sensor. B) Usage example of the head mask

3.7 COMSOL Simulation Results

Although we generally carried out experimental studies for the sensor design, we also wanted to see how the sensor behaves in the simulation environment. For simulation, we used a popular software called COMSOL Multiphysics, with which we can simulate many different physical models. The obtained result with version 5.4 is given in the figure below. The model was created using the optimum dimensions of the sensor. There are 20 fingers in total, 10 of which are on one side and 10 on the other.

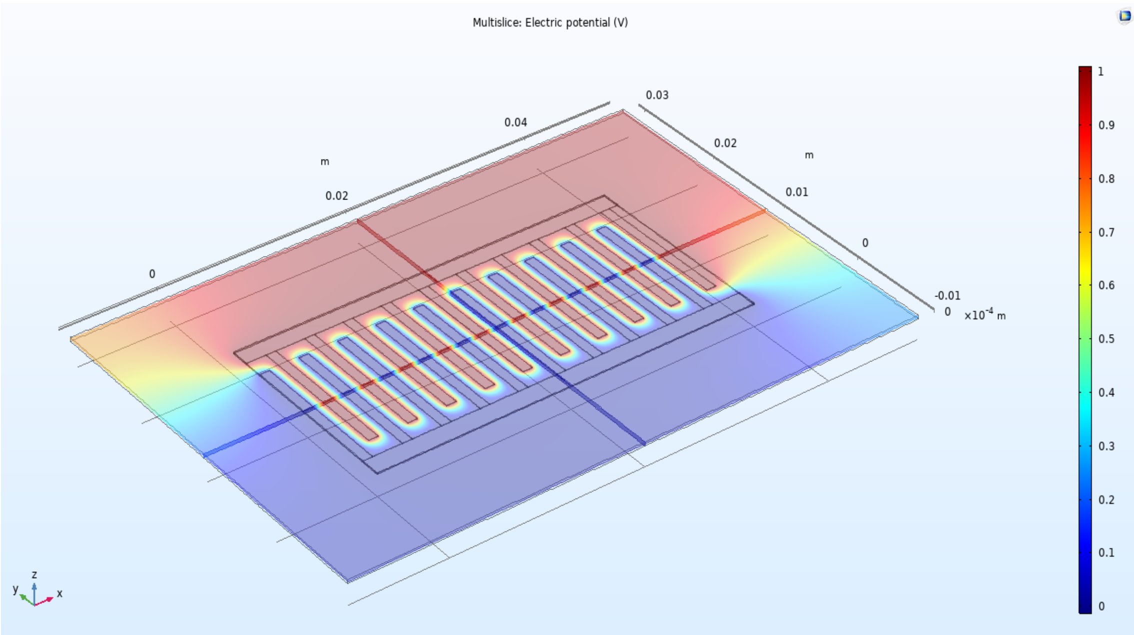

Figure 3.33 COMSOL simulation result of the sensor.

As seen in the figure, while one side of the sensor has a positive voltage, the other side can remain as ground. The points where the two sides meet, that is, between the fingers, have voltage values between positive and ground. So, the sensor can act as a capacitor. As the paper swells with moisture, distances of 0.1 cm between the fingers change, which causes a change in the capacitance value. Increasing the moisture in the paper has an effect to increase the capacitance value. When the sensor paper does not contain moisture, the capacitance value of the sensor returns to the initial value. Therefore, it can be concluded that the sensor can detect moisture changes. Respiration monitoring is possible by detecting the moisture content in human breath.

3.8 Cost Analysis

Cost analysis is the last subject to be discussed in this chapter. A general cost calculation has been made to produce the sensor and given in a table. The sensor has advantages in terms of sensitivity, speed, and easy production, its low cost is another important feature. The cost calculation is given in two tables. While the first table shows the cost required to manufacture a sensor, the second table calculates the cost of other materials needed to operate the sensor. According to the tables below, the production of the sensor can be realized as 7.61 TL, while its operation and proper use require 155.37 TL. A system that can detect and monitor respiration can be installed for a total of 163 TL. The prices given in the tables are the internet prices of products such as weighing paper, 2.5B pencil, PMMA stencil, Arduino Nano, cables, and filament, and these prices may change over time. Prices in the tables are the cheapest that can be found. The used stencil is also included in the price, but the cost for a sensor is lower since this stencil can be used over and over once it is produced.

Table 3.1 Cost required for the production of the sensor

Name of the product	Quantity	Price	Used	Total
Weighing Paper	250 pieces	366.82 TL	0.1 piece	1.47 TL
2.5B Pencil	12 pieces	58.90 TL	1 piece	4.9 TL
PMMA Stencil	1 piece 5x5 cm	1.24 TL	1 piece	1.24 TL
				7.61 TL

Table 3.2 Cost required for operation and proper use of the sensor

Name of the product	Quantity	Price	Used	Total
Arduino Nano	1 piece	133.34 TL	1 piece	133,34 TL
Cables	1x25 meter	49.32 TL	0.4 m	0.79 TL
Filament	1 kg	212.4 TL	0.1 kg	21.24 TL
				155.37 TL

Chapter 4

RESULTS

This section presents the experimental results of respiratory monitoring using the paper-based moisture sensor. First, the obtained results with and without a face mask are given, and then the experimental results on different subjects are presented.

4.1 Results of the Paper-Based Moisture Sensor

In the literature, it was reported that respiration monitoring can be done by using a face mask [4]. We have also achieved some results using a face mask. For this purpose, subjects were asked to breathe in a certain method toward the sensor. The mentioned method is as follows: It was waited for 10 seconds to observe the base value of the sensor, then subjects were asked to breathe normally for 1 minute and to start the deep breathing for another 1 minute, then after waiting for 1 minute, subjects were asked to breathe randomly for the last 1 minute. According to the results we obtained using this method, the difference between normal breathing and deep breathing disappeared in addition, the difference between mouth breathing and nose breathing disappeared due to the face mask. Moreover, the peak points in the figure no longer correspond to the time the subject breathes, instead, after 2 seconds, the sensor detects breathing. Also, the sensor could not return to its initial value during the 1-minute waiting period. Therefore, although the use of face masks can trap moisture in a certain place and achieve higher capacitance values, it is not preferred because it eliminates the difference between normal breathing and deep breathing and partially prevents its stable operation. The data obtained using the face mask are given in the figures below.

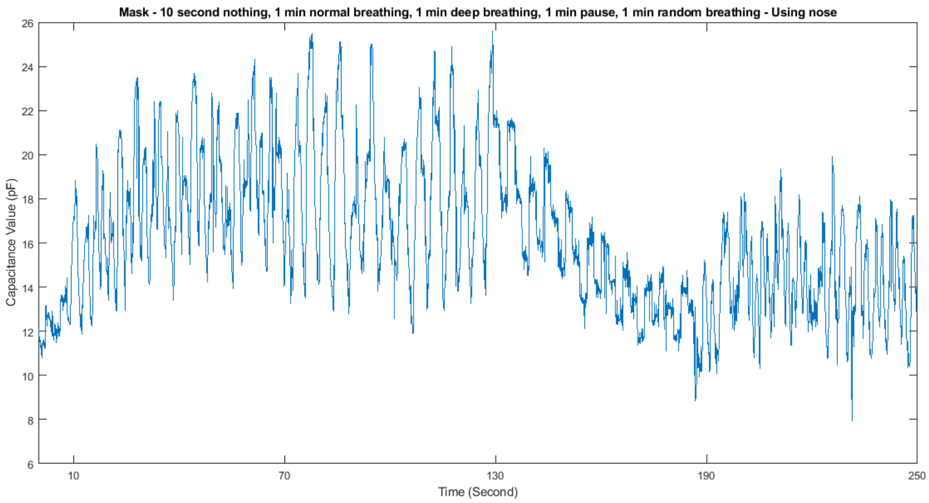

Figure 4.1 The experimental result of the sensor with the face mask using nose.

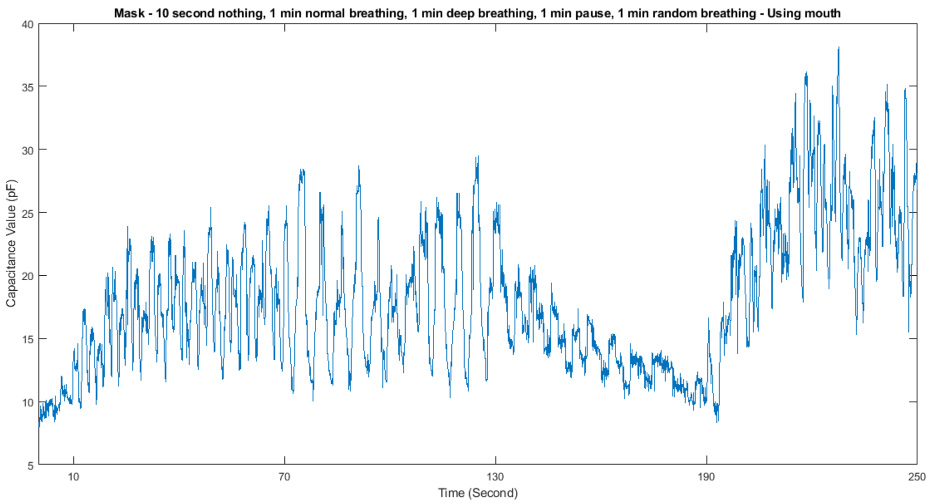

Figure 4.2 The experimental result of the sensor with the face mask using mouth.

We wanted to establish a standard method for the measurements that we performed using the moisture sensors. For this purpose, we used the same breathing method that we used for measurements with the face mask. Also, for this process, the distance between the sensor and the subject was fixed and determined as 6.5 cm. During the measurements, the subject only breathed through his mouth. With these conditions, we were finally able to make respiration monitoring. Furthermore, the sensor gave different responses to normal breathing and deep breathing. This means that the sensor can distinguish between deep and normal breathing. The obtained data from a subject using the sensor is given below.

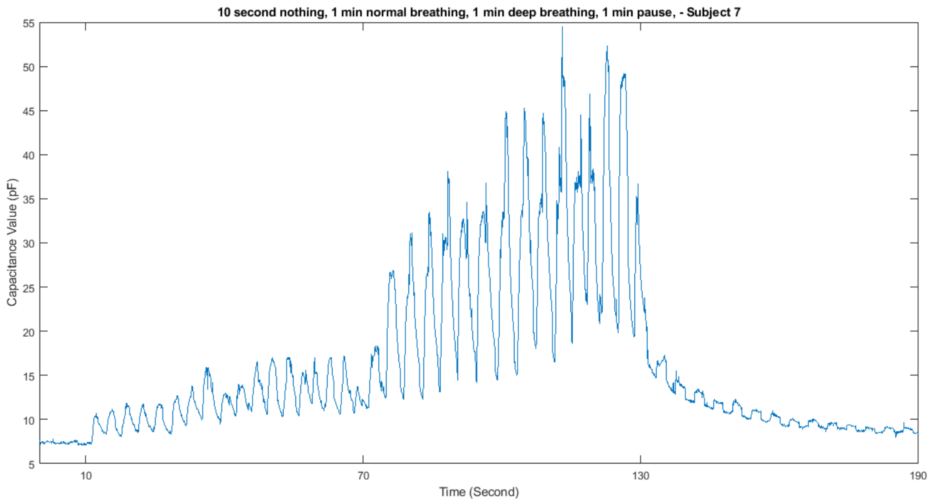

Figure 4.3 The experimental result of the sensor without the mask using mouth.

4.2 Experimental Results on Different Subjects

In order to investigate the capabilities of the sensor, we conducted experiments on three different groups, healthy/non-smokers, smokers and patients. The aim here is to test whether the sensor detects significant differences between these three groups. In such a case, the low-cost moisture sensor can give us information about the health status of the user. First, we collected samples from healthy/non-smokers, then from smokers and finally from patients. The sample collection procedure is the same for all subjects. We measured the base value of the sensor for the first 10 seconds, then asked the subjects to breathe normally for 1 minute, then we asked the subjects to breathe deeply for 1 minute, and finally, we observed how the sensor behaved for 1 minute. All these processes took 3 minutes and 10 seconds in total. We collected samples from a total of 84 people, 30 from non-smokers/healthy, 30 from smokers and 24 from patients. In order to analyze the results statistically, we generated various parameters. Each parameter has a deep/normal ratio. The meaning of these ratios is shown with mathematical formulas. These parameters and their meanings are as follows:

Normal: The data obtained from normal respiration. It can be seen in every parameter.

Deep: The data obtained from deep respiration. It can be seen in every parameter.

Deep/Normal: It is found by dividing the data obtained from deep respiration by the data obtained from normal respiration. It can be seen in every parameter.

Area: It is found by multiplying the base value of the sensor by the measurement time (usually 60) and subtracting the result from the area below the corresponding breath (deep or normal). It is shown visually in the figure below. After this process, the upper area, the blue part, in the figure is obtained.

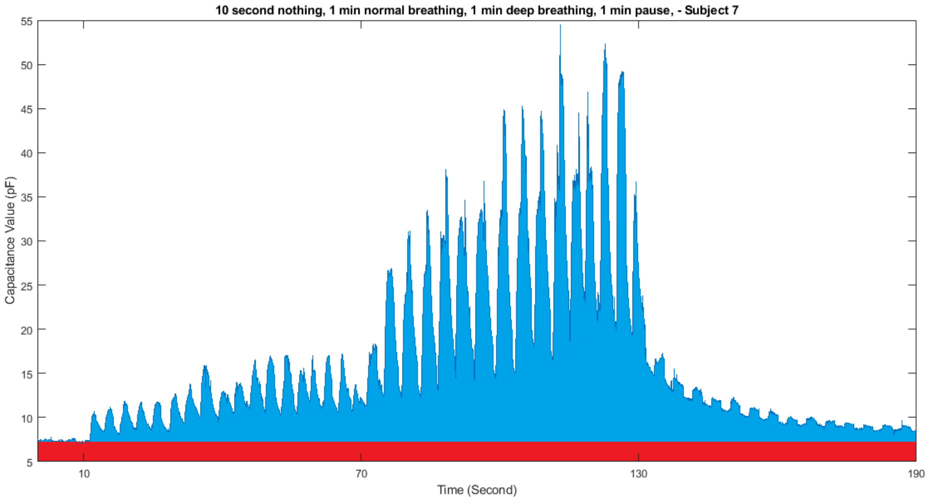

Figure 4.4 The calculation method of the area parameter.

$$\frac{D}{N} = \frac{The\ area\ under\ the\ deep\ breathing - 60 * sensor's\ base\ value}{The\ area\ under\ the\ normal\ breathing - 60 * sensor's\ base\ value} \quad (2)$$

Respiration rate: It shows the number of breaths taken during the corresponding breath (deep or normal). Deep/Normal ratio can be calculated as follows:

$$\frac{D}{N} = \frac{Deep\ respiration\ rate}{Normal\ respiration\ rate} \quad (3)$$

Max to min = It shows the maximum and minimum values of the sensor during the measurement. Normal means normal breathing maximum and minimum values. Deep means deep breathing maximum and minimum values. Deep/Normal ratio can be found as follows:

$$\frac{D}{N} = \frac{Deep\ breathing\ the\ max.\ value - Deep\ breathing\ the\ min.value}{Normal\ breathing\ the\ max.value - Normal\ breathing\ the\ min.value} \quad (4)$$

Avg. MtM. = It shows the average maximum-minimum difference over the corresponding breath (deep or normal). The mathematical formula is as follows:

$$\frac{D}{N}=\frac{\dfrac{(1.deep\ max.-1.deep\ min.)+(\ 2.deep\ max.-2.deep\ min.)+\cdots}{Res\ Rate\ Deep}}{\dfrac{(1.norm.max.-1.norm.min.)+(\ 2.norm.max.-2.norm.min.)+\cdots}{Res\ Rate\ Normal}} \quad (5)$$

The obtained results using these parameters are given in the tables below. First, the results of the non-smokers/healthy group, then the results of the smokers group and lastly the results of patients group are given.

Table 4.1 Experimental results of the Non-Smokers/Healthy group

Non-Smokers Healthy	Area (pF*s) Normal	Deep	Deep / Normal (pF*s)/ (pF*s)	Res. Rate (BPM) Nor – Deep		Deep/ Normal (BPM)/ (BPM)	Max. to Min (pF) Norm – Deep		Deep/ Normal (pF)/ (pF)	Avg. MtM (pF) Norm – Deep		Deep/ Normal (pF)/ (pF)
Subject NS - 1	100.9	193.1	1.91	19	10	0.53	12-8	20-8	3.00	2.43	6.53	2.69
Subject NS - 2	150.5	434.6	2.89	19	15	0.79	10-6	26-6	5.00	1.82	9.55	5.25
Subject NS - 3	117.4	276.9	2.34	17	13	0.76	11-7	17-7	2.50	1.62	4.97	3.06
Subject NS - 4	134.1	216.0	1.61	24	26	1.08	12-7	13-7	1.20	1.13	2.13	1.89
Subject NS - 5	84.5	124.7	1.48	23	28	1.22	8-6	9-6	1.50	1.20	1.68	1.40
Subject NS - 6	48.3	483.9	10.00	29	24	0.83	11-8	27-8	6.33	1.07	9.11	8.51
Subject NS - 7	95.2	196.1	2.06	13	7	0.54	8-6	11-6	2.50	0.95	3.67	3.86
Subject NS - 8	273.5	1184.5	4.33	18	16	0.89	17-7	54-7	4.70	4.41	23.7	5.37
Subject NS - 9	123.9	130.0	1.05	19	21	1.11	11-7	10-7	0.80	1.70	1.84	1.08
Subject NS - 10	63.9	126.4	1.98	21	20	0.95	9-7	12-7	2.50	1.01	2.18	2.15
Subject NS - 11	364.9	850.9	2.33	16	10	0.63	19-5	44-5	2.79	6.90	21.8	3.16
Subject NS - 12	78.3	587.5	7.50	15	10	0.67	11-8	37-8	11.17	1.64	16.3	9.91
Subject NS - 13	90.9	368.2	4.05	30	11	0.37	11-8	23-8	4.95	1.21	8.06	6.64
Subject NS - 14	87.8	238.3	2.71	18	19	1.06	9-6	14-6	2.91	0.93	4.21	4.52
Subject NS - 15	493.4	822.1	1.67	20	13	0.65	29-8	42-8	1.63	7.51	19.6	2.61
Subject NS - 16	270.9	606.1	2.24	18	16	0.89	18-6	30-6	2.03	4.18	12.1	2.91
Subject NS - 17	119.8	147.6	1.23	27	26	0.96	10-7	11-7	1.13	1.47	1.74	1.19
Subject NS - 18	57.4	98.11	1.71	19	21	1.11	9-8	10-8	2.02	0.45	1.04	2.30
Subject NS - 19	55.5	125.8	2.27	17	7	0.41	10-8	12-8	2.32	0.77	2.14	2.78
Subject NS - 20	101.7	322.2	3.17	22	18	0.82	11-8	19-8	3.60	1.36	5.03	3.69
Subject NS - 21	137.5	231.9	1.69	25	20	0.80	12-8	17-8	2.03	1.35	3.70	2.75
Subject NS - 22	96.7	127.6	1.32	15	12	0.80	10-7	12-7	1.77	1.59	2.65	1.66
Subject NS - 23	348.7	1331.7	3.82	21	14	0.67	22-6	77-6	4.35	8.16	44.1	5.40
Subject NS - 24	175.2	572.8	3.27	26	21	0.81	13-7	37-7	4.83	2.08	9.68	4.66
Subject NS - 25	79.3	164.5	2.07	17	14	0.82	13-8	15-8	1.38	1.78	3.11	1.74
Subject NS - 26	1180.8	3349.8	2.84	20	16	0.80	88-7	181-7	2.15	26.3	122	4.64
Subject NS - 27	109.9	193.2	1.76	19	15	0.79	10-7	13-7	2.21	1.35	3.05	2.27
Subject NS - 28	72.6	93.0	1.28	18	12	0.67	11-8	13-8	1.42	1.44	2.01	1.40
Subject NS - 29	69.7	108.5	1.56	15	11	0.73	10-8	11-8	1.97	1.24	2.37	1.91
Subject NS - 30	135.1	293.0	2.17	22	18	0.82	11-5	24-5	3.29	3.97	9.35	2.35

We also collected data from smokers using the same methods and parameters. The results are as follows.

Table 4.2 Experimental results of the Smokers group

Smokers	Area (pF*s)		Deep / Normal (pF*s)/ (pF*s)	Res. Rate (BPM)		Deep/ Normal (BPM)/ (BPM)	Max. to Min (pF)		Deep/ Normal (pF)/ (pF)	Avg. MtM (pF)		Deep/ Normal (pF)/ (pF)
	Normal	Deep		Nor	Deep		Norm	Deep		Norm	Deep	
Subject S - 1	234.5	615.1	2.62	19	19	1.00	14-5	30-5	2.78	3.53	13.20	3.74
Subject S - 2	65.1	108.4	1.66	23	19	0.83	10-8	11-8	1.67	0.97	1.57	1.62
Subject S - 3	133.0	356.1	2.68	13	13	1.00	15-7	20-7	1.62	3.02	8.07	2.67
Subject S - 4	249.8	339.7	1.36	28	26	0.93	17-6	22-7	1.46	4.07	8.44	2.07
Subject S - 5	71.8	157.3	2.19	14	15	1.07	10-7	12-7	1.79	1.36	2.58	1.89
Subject S - 6	85.8	241.5	2.81	33	32	0.97	11-8	16-8	2.66	1.17	3.46	2.96
Subject S - 7	86.2	112.4	1.30	26	16	0.62	10-8	11-8	1.61	0.90	1.83	2.04
Subject S - 8	76.3	196.1	2.57	26	19	0.73	11-8	14-8	2.01	1.04	3.52	3.39
Subject S - 9	116.1	136.0	1.17	23	14	0.61	10-7	11-7	1.15	1.10	1.65	1.50
Subject S - 10	137.64	217.4	1.58	24	24	1	12-7	13-7	1.26	1.93	2.86	1.48
Subject S - 11	79.9	168.0	2.10	18	8	0.44	10-8	13-8	2.53	1.20	3.29	2.75
Subject S - 12	108.9	216.8	1.99	17	9	0.53	12-7	13-7	1.21	1.90	3.69	1.94
Subject S - 13	72.1	176.9	2.45	20	14	0.70	10-8	14-8	3.29	1.09	3.37	3.09
Subject S - 14	200.8	234.6	1.17	13	8	0.62	12-6	14-6	1.36	2.72	4.01	1.48
Subject S - 15	95.5	203.9	2.13	21	16	0.76	9-7	14-7	2.82	1.59	3.64	2.29
Subject S - 16	186.2	193.3	1.04	27	18	0.67	14-7	16-7	1.48	2.73	4.38	1.60
Subject S - 17	126.5	205.0	1.62	21	18	0.86	13-8	15-8	1.38	2.05	3.65	1.78
Subject S - 18	82.0	192.9	2.35	22	15	0.68	10-8	14-8	2.60	1.15	3.26	2.84
Subject S - 19	94.2	248.8	2.64	25	16	0.64	12-8	20-8	3.07	1.19	4.58	3.85
Subject S - 20	177.8	299.0	1.68	17	14	0.82	14-6	15-6	1.20	2.64	5.34	2.02
Subject S - 21	172.1	201.9	1.17	21	20	0.95	11-7	12-7	1.22	2.02	2.61	1.29
Subject S - 22	62.5	145.1	2.32	19	12	0.63	9-7	20-7	8.06	1.11	4.31	3.88
Subject S - 23	65.4	93.7	1.43	13	13	1.00	10-8	11-8	1.38	1.25	1.91	1.53
Subject S - 24	104.7	250.6	2.39	21	19	0.90	10-7	15-7	2.94	1.33	3.26	2.46
Subject S - 25	100.9	161.5	1.60	24	13	0.54	11-8	14-8	2.02	1.86	4.06	2.18
Subject S - 26	88.6	197.2	2.22	21	18	0.86	12-8	17-8	2.32	1.94	4.10	2.12
Subject S - 27	107.1	169.0	1.58	21	17	0.81	10-7	12-7	1.43	1.33	2.36	1.77
Subject S - 28	122.0	154.5	1.27	15	12	0.80	11-7	13-7	1.40	2.01	3.03	1.50
Subject S - 29	104.4	266.9	2.56	12	9	0.75	13-7	17-7	1.71	2.02	5.27	2.61
Subject S - 30	107.7	200.3	1.86	20	16	0.80	10-7	13-7	1.85	1.66	2.96	1.78

Lastly, the patients group is given below. We also used same parameters for this group. There are 24 subjects for the patients group. This group consists of patients with pneumonia and COPD.

Table 4.3 Experimental results of the Patients group

Patients	Area (pF*s) Normal	Deep	Deep / Normal (pF*s)/ (pF*s)	Res. Rate (BPM) Nor – Deep		Deep/ Normal (BPM)/ (BPM)	Max. to Min (pF) Norm – Deep		Deep/ Normal (pF)/ (pF)	Avg. MtM (pF) Norm – Deep		Deep/ Normal (pF)/ (pF)
Subject P - 1	73.1	168.1	2.29	16	17	1.06	10-6	17-6	2.48	1.28	3.23	2.51
Subject P - 2	600.0	1518	2.52	33	22	0.66	36-8	65-8	2.02	8.63	32.5	3.76
Subject P - 3	99.1	219.6	2.21	26	20	0.77	10-6	13-6	1.82	0.82	2.04	2.49
Subject P - 4	88.4	131.9	1.49	14	15	1.07	10-3	12-3	1.35	1.51	2.20	1.46
Subject P - 5	133.5	217.1	1.62	30	23	0.76	11-9	14-9	2.21	0.63	1.43	2.26
Subject P - 6	47.4	60.2	1.27	14	11	0.78	10-6	9-6	0.88	1.51	1.46	0.96
Subject P - 7	32.0	101.5	3.16	20	17	0.85	9-7	12-7	1.91	0.89	1.91	2.13
Subject P - 8	97.0	171.7	1.77	26	21	0.81	12-8	13-8	1.21	0.81	1.75	2.17
Subject P - 9	95.1	150.7	1.58	20	17	0.85	12-7	14-7	1.30	1.03	1.57	1.52
Subject P - 10	9.6	51.1	5.28	17	17	1.00	9-7	9-7	1.77	0.68	0.85	1.24
Subject P - 11	47.1	212.1	4.49	13	11	0.84	11-3	19-3	2.08	0.76	1.79	2.36
Subject P - 12	89.4	195.5	2.18	23	15	0.65	11-8	15-8	2.84	1.24	4.05	3.26
Subject P - 13	95.6	310.4	3.24	45	34	0.75	12-9	25-9	5.46	0.63	3.51	5.52
Subject P - 14	93.1	129.9	1.39	30	19	0.63	10-7	11-7	1.29	0.81	1.21	1.48
Subject P - 15	176.4	236.7	1.34	17	14	0.82	17-5	15-5	0.85	1.91	2.77	1.44
Subject P - 16	43.8	133.6	3.04	18	15	0.83	11-5	16-5	2.02	0.77	2.78	3.57
Subject P - 17	203.0	335.8	1.65	36	34	0.94	14-6	17-6	1.38	1.88	3.48	1.84
Subject P - 18	104.2	184.3	1.76	17	13	0.76	10-7	11-7	1.39	1.24	2.03	1.63
Subject P - 19	44.7	149.1	3.33	31	27	0.87	10-8	12-8	2.06	0.94	1.58	1.68
Subject P - 20	336.5	354.1	1.05	22	14	0.63	24-5	27-5	1.18	5.26	7.76	1.47
Subject P - 21	973.4	1050	1.07	21	12	0.57	55-7	65-7	1.21	12.1	16.8	1.38
Subject P - 22	103.7	151.4	1.45	17	14	0.82	12-6	13-6	1.06	1.61	2.22	1.37
Subject P - 23	135.1	354.0	2.61	18	16	0.88	16-8	21-8	1.61	2.46	4.26	1.72
Subject P - 24	159	206.9	1.30	19	15	0.78	14-7	15-7	1.16	1.62	2.34	1.44

For statistical evaluation, we investigated whether the data are normally distributed for each parameter. We used the Kolmogorov-Smirnov Test to determine the distributions of three data sets. According to this test, if the p-value is greater than 0.05, data are normally distributed. We performed this test and obtained p-values for each parameter. Distributions of data for the non-smokers/healthy group are given below as a table.

Table 4.4 The Kolmogorov-Smirnov Test results of the Non-Smokers/Healthy group

Parameter (Non-Smokers/Healthy group)	p-value	Distribution
Deep Area/Normal area	0.0541	Normal
Deep Res. Rate/Normal Res. Rate	0.5681	Normal
Deep Max to Min /Normal Max to Min	0.1584	Normal
Deep Avg. MtM /Normal Avg. MtM	0.2008	Normal

Since the p-values for all parameters are greater than 0.05, it is determined that the data are normally distributed for all parameters. Therefore, the t-test will be used for statistical analysis of the non-smokers/healthy group. We repeated the same procedure for the smokers group and determined the distributions of parameters. Again, we used the Kolmogorov-Smirnov Test. Distributions of data for the smokers group are given in the table below.

Table 4.5 The Kolmogorov-Smirnov Test results of the Smokers group

Parameter	p-value	Distribution
Deep Area/Normal area	0.5797	Normal
Deep Res. Rate/Normal Res. Rate	0.9754	Normal
Deep Max to Min /Normal Max to Min	0.8000	Normal
Deep Avg. MtM /Normal Avg. MtM	0.4762	Normal

Since the p-values for all parameters are greater than 0.05, it is determined that the data of smokers are also normally distributed. Again, for statistical analysis we need to use t test for the smokers group. Lastly, we have the patients group. Using the same test we found distributions of data for the patients group. The result is given below as a table.

Table 4.6 The Kolmogorov-Smirnov Test results of the Patients group

Parameter	p-value	Distribution
Deep Area/Normal area	0.2416	Normal
Deep Res. Rate/Normal Res. Rate	0.8565	Normal
Deep Max to Min /Normal Max to Min	0.2452	Normal
Deep Avg. MtM /Normal Avg. MtM	0.2976	Normal

So far all parameters are normally distributed. This means we need to use the t-test to compare these three groups, so we used a t-test for all parameters separately. First we compared smokers and non-smokers/healthy groups. According to the result we reached by using the two-tailed t-test, only the respiration rate D/N ratio does not give a statistically significant result. It seems that respiration rate is a parameter that cannot determine a significant difference between smokers and non-smokers/healthy groups. Other than the respiration rate D/N ratio, we found statistically significant differences for the rest of the parameters since the p-values are less than 0.05. According to this result, we can say that the moisture sensor is a sensor that can detect the differences between these two groups. Thus, the sensor not only monitors respiration but also provides information about the health status of users. Considering that the production of the sensor is extremely easy, and the cost is quite low, and also it provides information about the health status of the users, we can say that the sensor is a great candidate for further development. While the p-value is less than 0.05 for other parameters, the p-value is less than 0.01 for the Average Maximum to Minimum D/N Ratio parameter. The p-value says whether the data obtained is by chance. A p-value of 0.05 means that the difference between the data is obtained by chance, with a 5% probability. In other words, there is a 95% probability that the difference is not achieved by chance. Therefore, for the Average Maximum to Minimum D/N Ratio parameter, the differences between the two groups are not obtained by chance with a probability of more than 99%. Consequently, with this parameter, the sensor reveals highly statistically significant data. The obtained results from the unpaired t-test are given in the table below.

Table 4.7 The t-test result of Non-Smokers/Healthy and Smokers groups

<table>
<tr><th>Parameter</th><th>Group</th><th>Non-Smokers</th><th>Smokers</th><th>Sample Size</th></tr>
<tr><td rowspan="5">Area D/N Ratio (pF*s)</td><td>Mean</td><td>2.6770</td><td>1.9170</td><td rowspan="5">30-30</td></tr>
<tr><td>Standard Deviation</td><td>1.8779</td><td>0.5470</td></tr>
<tr><td>Standard error of the mean</td><td>0.3429</td><td>0.0999</td></tr>
<tr><td>t-value</td><td colspan="2">2.1283</td></tr>
<tr><td>p-value</td><td colspan="2">0.0376</td></tr>
<tr><td rowspan="5">Respiration Rate D/N Ratio (BPM)</td><td>Mean</td><td>0.7993</td><td>0.7840</td><td rowspan="5">30-30</td></tr>
<tr><td>Standard Deviation</td><td>0.2006</td><td>0.1648</td></tr>
<tr><td>Standard error of the mean</td><td>0.0366</td><td>0.0301</td></tr>
<tr><td>t-value</td><td colspan="2">0.3234</td></tr>
<tr><td>p-value</td><td colspan="2">0.7475</td></tr>
<tr><td rowspan="5">Maximum to Minimum D/N Ratio (pF)</td><td>Mean</td><td>2.9993</td><td>2.1093</td><td rowspan="5">30-30</td></tr>
<tr><td>Standard Deviation</td><td>2.0594</td><td>1.2951</td></tr>
<tr><td>Standard error of the mean</td><td>0.3760</td><td>0.2365</td></tr>
<tr><td>t-value</td><td colspan="2">2.0037</td></tr>
<tr><td>p-value</td><td colspan="2">0.0498</td></tr>
<tr><td rowspan="5">Average Maximum to Minimum D/N Ratio (pF)</td><td>Mean</td><td>3.4583</td><td>2.2707</td><td rowspan="5">30-30</td></tr>
<tr><td>Standard Deviation</td><td>2.1247</td><td>0.7544</td></tr>
<tr><td>Standard error of the mean</td><td>0.3879</td><td>0.1377</td></tr>
<tr><td>t-value</td><td colspan="2">2.8852</td></tr>
<tr><td>p-value</td><td colspan="2">0.0055</td></tr>
</table>

We applied t-test for non-smokers/healthy and patients groups. This time, no statistical difference was found for area D/N ratio and respiration rate D/N ratio parameters. Instead, the maximum to minimum D/N ratio parameter gives great statistical results because the p-value is less than 0.01. The p value of the last parameter is also less than 0.01 as in the non-smokers/healthy and smokers group. Thus, we were able to show statistically that the sensor we produced can distinguish between patients and healthy people so we can make comment about the health status of users.

Table 4.8 The t-test result of Non-Smokers/Healthy and Patients groups

Parameter	Group	Healthy	Patients	Sample Size
Area D/N Ratio (pF*s)	Mean	2.6770	2.2121	30-24
	Standard Deviation	1.8779	1.0861	
	Standard error of the mean	0.3429	0.2217	
	t-value	1.0762		
	p-value	**0.2868**		
Respiration Rate D/N Ratio (BPM)	Mean	0.7993	0.8075	30-24
	Standard Deviation	0.2006	0.1283	
	Standard error of the mean	0.0366	0.0262	
	t-value	0.1729		
	p-value	**0.8634**		
Maximum to Minimum D/N Ratio (pF)	Mean	2.9993	1.7725	30-24
	Standard Deviation	2.0594	0.9378	
	Standard error of the mean	0.3760	0.1914	
	t-value	2.6993		
	p-value	**0.0094**		
Average Maximum to Minimum D/N Ratio (pF)	Mean	3.4583	2.1108	30-24
	Standard Deviation	2.1247	1.0315	
	Standard error of the mean	0.3879	0.2106	
	t-value	2.8464		
	p-value	**0.0063**		

Finally, we compared patients and smokers using t test. We did not find any statistical difference as expected. The p value for all parameters is greater than 0.05. That means there are no statistically significant differences. In this way, we have shown once again that the sensor does not randomly detect the differences between the non-smokers/healthy and the smokers-patients groups. Another conclusion to be drawn from this results is that smoking has as bad consequences as diseases such as pneumonia or COPD. The results are given in the table below.

Table 4.9 The t-test result of Patients and Smokers groups

Parameter	Group	Patients	Smokers	Sample Size
Area D/N Ratio (pF*s)	Mean	2.2121	1.9170	24-30
	Standard Deviation	1.0861	0.5470	
	Standard error of the mean	0.2217	0.0999	
	t-value	1.2985		
	p-value	**0.1998**		
Respiration Rate D/N Ratio (BPM)	Mean	0.8075	0.7840	24-30
	Standard Deviation	0.1283	0.1648	
	Standard error of the mean	0.0262	0.0301	
	t-value	0.5729		
	p-value	**0.5692**		
Maximum to Minimum D/N Ratio (pF)	Mean	1.7725	2.1093	24-30
	Standard Deviation	0.9378	1.2951	
	Standard error of the mean	0.1914	0.2365	
	t-value	1.0688		
	p-value	**0.2901**		
Average Maximum to Minimum D/N Ratio (pF)	Mean	2.1108	2.2707	24-30
	Standard Deviation	1.0315	0.7544	
	Standard error of the mean	0.2106	0.1377	
	t-value	0.6575		
	p-value	**0.5138**		

Chapter 5

Conclusions and Future Prospects

5.1 Conclusions

In this thesis, we produced a paper-based, low-cost, easy-to-manufacture, fast-response, high-accuracy, and easy-to-use moisture sensor. The sensor is ideal for respiration monitoring as the sensor can still maintain its properties in long-term use. Contrary to other studies in the literature, respiratory monitoring does not disturb the users since face masks are not required for the use of the moisture sensor.

We first determined the paper to be used for the sensor production. The paper is the most important material for the characteristics of the sensor. For this reason, we worked for a long time to find the right paper. In the end, we decided to use ISOLAB weighing paper. This paper is a paper that can absorb the moisture around it rapidly and give it back to the outside environment.

After determining the paper, we started researching the pencil to be used. Even if it is not as much as paper, the pencil is also important for the sensor to work properly. After several trials, we had two candidates. One was a pencil with a hardness of 2.5B, while the other was a pencil with a hardness of 9B. In the end, the obtained results from these two pencils were very close to each other, so both could be used. We have decided to use the pencil with a hardness of 2.5B.

Another important point is the pattern to be drawn on the paper. If we didn't use the right pattern, the right paper and pencil wouldn't matter. Because the sensor could not detect respiration without the correct pattern. We tried different designs and we saw from the results that the interdigitated design, which was used in the literature before, was very suitable for the sensor. As a result, we decided to use interdigitated pattern. We also made the stencil out of PMMA sheets. In addition, we did not have any problems with PMMA stencil production as we also had easy access to laser printers.

After doing various experiments such as painting the back of the sensor and testing the durability, we designed an electronic circuit for measurements. To use it as a wearable system, we used an Arduino Nano microcontroller. We have modified the codes we use so that the measurements can be made properly.

Then we printed mechanical parts from 3D printers to get standard measurements. Thus, we kept the distance constant and subjects only breathed through their mouths. Lastly, we minimized the errors caused by external factors.

In conclusion, we have produced a system that can detect and monitor respiration. This system can be preferred due to its low cost and high accuracy. Another advantage of the system is that the sensor can detect the respiration rate as well as the difference between normal breathing and deep breathing. By not using a mask in the system, we have enabled it to be used by people who have respiratory distress. Lastly, using the sensor we can comment on the health status of users.

5.2 Societal Impact and Contribution to Global Sustainability

The moisture sensor can detect respiration with high accuracy. Since the sensor is paper based, as mentioned before, it is very cheap to manufacture. This makes the sensor sustainable. Moreover, the cost required for respiration monitoring will be reduced. Since the sensor is controlled by the Arduino microcontroller, the labor required for respiratory monitoring will also decrease, which will have an extra effect on the cost. Thus, this system that we have designed can be preferred individually as it is easy to use and can also be used by institutions such as hospitals. Therefore, this system that we have produced can be used in patient follow-up. In this way, patients can be followed instantly by using an application, as well as reducing the workload on hospital staff. In addition, thanks to its fast and easy production in cases such as the COVID-19 epidemic, it can enable society to recover from these epidemic situations with less damage. These would provide a positive social impact. Moreover, since the other materials used for the sensor are low-cost, they can be produced and set up in a very short time, which makes it very easy for hospital staff to monitor respiratory under epidemic conditions. This system can also be used in the detection of some diseases such as sleep apnea, it can be used not only for respiration monitoring but also for disease detection.

This system is economically sustainable since the paper, which is the main material of the sensor, can be recycled, and the sensor maintains its properties even if it is used repeatedly. Since the used materials in the production of the sensor are not harmful to nature, it is also a sensor with a positive environmental impact. Considering all these factors, it is a positive study in terms of both social and environmental effects.

5.3 Future Prospects

One of the studies to be done within the scope of future prospect for this study may be to change the production method of the sensor. During the experimental study, we used a stencil and a pencil to meet our sensor needs and drew the necessary patterns by hand. If the sensor is to be used widely by more people, it is obvious that one person cannot produce all the sensors. Instead of this production method, an automation-based production method can be used by using machines. Thus, a large number of sensors could possibly be manufactured at a lower cost and the sensor could be widely used for respiration monitoring.

Another issue may be the development of artificial intelligence that can record the data obtained from user's breaths and compare them with historical data. Thus, if an abnormality is detected in the breathing of the user for a while, some possible diseases can be detected or some situations that may lead to death can be prevented by warning the user. An example of one of these situations is cardiac arrest. Cardiac arrest causes respiratory abnormalities [78]. Therefore, in such a system, in addition to life-threatening conditions such as cardiac arrest, some diseases can be detected.

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APPENDIX

Appendix A: Arduino Code

```
const int OUT_PIN = A2;
const int IN_PIN = A0;
const int led = 13;
const float IN_STRAY_CAP_TO_GND = 24.48;
const float IN_CAP_TO_GND  = IN_STRAY_CAP_TO_GND;
const float R_PULLUP = 34.8;
const int MAX_ADC_VALUE = 1023;
int cap_old=0;
unsigned long previousMillis = 0;
const long interval = 500;

void setup(){
 pinMode(OUT_PIN, OUTPUT);
 pinMode(IN_PIN, OUTPUT);
 pinMode(led, OUTPUT);
 Serial.begin(9600);
}

void loop(){
  pinMode(IN_PIN, INPUT);
  digitalWrite(OUT_PIN, HIGH);
  int val = analogRead(IN_PIN);
  digitalWrite(OUT_PIN, LOW);

  if (val < 1000){
   pinMode(IN_PIN, OUTPUT);
   float capacitance = (float)val * IN_CAP_TO_GND / (float)(MAX_ADC_VALUE -
val);
   Serial.println(capacitance, 3);
   unsigned long currentMillis = millis();
```

```
    if((currentMillis - previousMillis >= interval))
    {
     previousMillis = currentMillis;

    if((capacitance > 4))
    {
     digitalWrite(led,HIGH);
    }
    else
    {
     digitalWrite(led,LOW);
    }
    }
    cap_old=capacitance;
   }
}
```

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