# Graphene Oxide-Based Resistive Humidity Sensor for Non-Contact Monitoring of Human Breathing Patterns

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## Abstract:

This work presents a simple, low-cost, and effective resistive humidity sensor based on graphene oxide for monitoring human breathing patterns in a non-contact manner. The sensor was fabricated by drop-casting a graphene oxide suspension onto a commercial Zensor TE100 platform with printed electrodes. A voltage divider circuit paired with an op-amp was used to convert the sensor's resistance changes into measurable voltage signals, which were recorded using MATLAB's analog input interface. The sensor reliably detected various breathing patterns—mouth and nasal breathing, single and continuous cycles, and both low- and high-intensity airflow. Results show that exhaled humidity significantly decreases the sensor's resistance, producing a clear signal during each breath. Notably, high-intensity exhalation led to baseline drift, likely due to deeper water molecule adsorption within the porous graphene oxide structure. This behavior highlights the material's sensitivity and reveals both opportunities and challenges for long-term respiration monitoring. Overall, the study demonstrates the potential of graphene oxide humidity sensors for wearable or ambient respiratory health monitoring applications.

***Keywords***: breathe analysis, humidity sensor, graphene oxide

## Introduction:

Humidity sensors have traditionally been used to monitor environmental conditions in areas such as agriculture, industry, and smart home systems. Recently, however, their role has expanded with the rise of wearable electronics and health monitoring technologies. This shift has led to growing interest in developing sensors tailored for applications involving the human body [1, 2].

These new applications are both diverse and innovative. For example, humidity sensors can non-invasively monitor breathing patterns such as rate, strength, and variation by detecting moisture in exhaled air. They can also pick up on humidity changes during speech, allowing for simple voice recognition. On the skin, these sensors help detect moisture levels, even though humans sense humidity indirectly through touch and temperature. Moreover, fingertip moisture can be used to operate devices via non-contact switches, and humidity sensors are now being used in diaper monitoring systems, some of which are already commercially available [2].

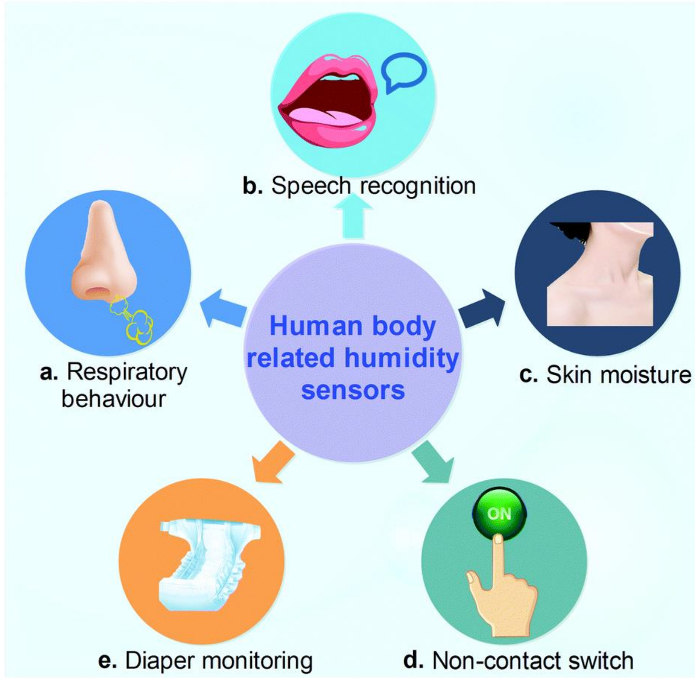


*Figure 1. Humidity and moisture form body [2]*

Compared to conventional environmental humidity sensors, those designed for use on the human body must meet a range of additional requirements. They need to be flexible, respond and recover quickly, and be completely safe for direct contact with the skin. To meet these demands, researchers have made significant progress in recent years. For example, to keep up with fast breathing rates, sensors with rapid response and recovery times have been developed. To ensure comfort and adaptability for wearable devices, flexible sensor designs have been introduced. Additionally, to improve safety and reduce production costs, eco-friendly and low-cost humidity sensors have been explored [3].

Still, turning these advancements into practical, real-world solutions is not without its challenges. Ongoing efforts are focused on improving the materials used for humidity sensing, optimizing sensor design, and enhancing signal processing and data analysis. Despite the progress, fully realizing the potential of humidity sensors for human body applications continues to require further innovation and refinement. The material used to sense humidity is the heart of the sensor, directly affecting how well it works for detecting moisture on the human body. Over the years, researchers have explored many different materials for this purpose. Some of the most common ones include cellulose-based paper, carbon-based materials, polymers, two dimensional materials beyond graphene, metal oxides, and various combinations of these. Each material has its own strengths, such as being flexible, sensitive, or safe for use on the skin, making them ideal for different types of wearable or body-related humidity sensors[4].

Resistance-based humidity sensors designed for human body applications and operating in direct-current (DC) mode have been widely studied, largely due to the simplicity of their signal processing circuits. Their straightforward design makes them easier to integrate into wearable and portable systems, which is especially important for real-time monitoring and low-power applications [2].

Graphene oxide is an excellent material for humidity sensing applications, particularly for use in sensors designed for the human body. Its unique structure, which includes oxygen functional groups, makes it highly sensitive to changes in moisture levels. The presence of these oxygen groups allows graphene oxide to interact with water molecules, leading to significant changes in its electrical conductivity in response to varying humidity. This makes it ideal for detecting even small fluctuations in moisture, which is crucial for applications like monitoring skin moisture or respiratory patterns. Additionally, graphene oxide is flexible, lightweight, and can be easily processed into thin films, making it well-suited for wearable sensors. Its biocompatibility and the ability to integrate with other materials further enhance its potential for safe, efficient, and responsive humidity sensors for human body-related applications [5].

In this paper, preliminary results about a resistive humidity senor based on graphene oxide for finding number of breathes per minutes. Local maxima in Sensor's signal can be used as number exhale and duration between two local minima related to duration of breath. Moreover, the baseline of sensor can give information about irregular breath pattern.

## Materials and method:

### Sensor fabrication:

To fabricate this sensor, a TE100 Zensor is used as both the substrate and the electrode platform. The TE100 contains three screen-printed electrodes: a reference electrode (silver), a working electrode (graphite), and a counter electrode (also graphite). First, the TE100 is rinsed with water and dried under a chemical hood to remove any contaminants from the electrode surfaces. Next, 200 µL of graphene oxide (4 mg/mL, purchased from Sigma-Aldrich) is ultrasonicated to ensure uniform dispersion. Finally, 50 µL of the solution is drop-cast onto the surface of the electrode and left to be dried for a day.

### Measurement circuit:

According to many studies, the resistance of graphene oxide decreases as humidity increases. To monitor this change, the resistance of the sensor is converted into a voltage signal using a simple voltage divider circuit. In this setup, the sensor is connected in series with a known resistor. As humidity changes, the resistance of the graphene oxide changes which in turn alters the voltage at the node between the resistor and sensor. This voltage variation corresponds to changes in humidity. The voltage divider is powered by a 5 V supply.

### Readout circuit:

The sensor exhibits a resistance greater than 10 MΩ under ambient conditions. While any resistor can technically be used in the voltage divider, a practical choice is one that is up to an order of magnitude smaller than the sensor resistance. It's important to note that in the idle state, DC biasing consumes minimal power less than 2.5 µW. Ideally, for optimal power matching, a resistor with the same resistance as the sensor would be used. This would enable configurations like a Wheatstone bridge for more precise measurements. However, in solution-processed sensor fabrication, achieving two identical sensors using drop-casting is challenging and complicates the design.

Considering these trade-offs, a 100 kΩ resistor was selected as the fixed series resistor. During exhalation, the voltage at the divider node can rise to approximately 0.2 V. To amplify this signal, a non-inverting amplifier based on the LF356 op-amp is used with gain of 11. The amplified output is then fed into a single-ended analog input channel of the NI USB-6008 data acquisition device for recording and analysis. Figure 2 shows the experimental setup for reading signal.

**Recording and process:**

For data recording, the Analog Input Recorder app in MATLAB is used. This app provides a graphical user interface (GUI) that allows users to easily configure recording settings, including sampling frequency, duration, and ADC window. For my test, 1000 sample per second, 1 minute, and -5 V to 5V is used. After data is collected, a low-pass filter is applied to the recorded signal to reduce noise and extract relevant features. The MATLAB code used for filtering is provided in the appendix of this paper.

**Test protocol:**

During testing, the sensor was positioned approximately 5–8 cm away from the nose or mouth. Various breathing patterns were performed, including single breaths, as well as fast and slow inhale–exhale cycles.

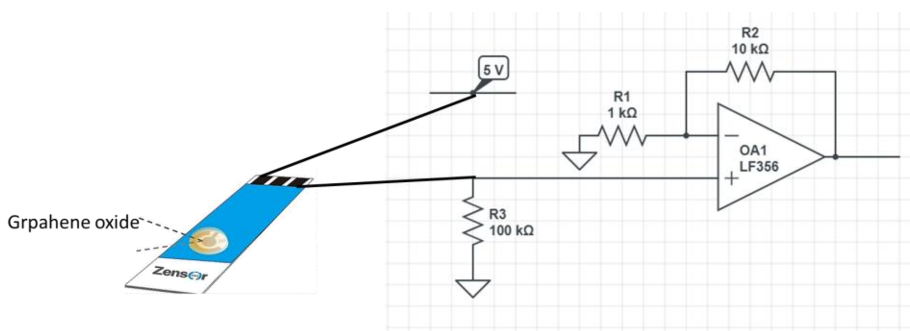


*Figure 2. Experimental setup*

## Results and discussion:

Graphene oxide exhibits excellent resistive humidity sensing properties, making it well-suited for monitoring breathing patterns. Its electrical resistance decreases with increasing humidity due to the absorption of water molecules by its oxygen-containing functional groups, which enhances ionic conductivity. During exhalation, the moisture in the breath increases the local humidity around the sensor, causing a noticeable drop in resistance. In contrast, during inhalation, the humidity decreases, leading to a rise in resistance [6]. This cyclic change produces a clear and repeatable signal that aligns with the breathing pattern. From an electrical output perspective, as the sensor resistance decreases during exhalation, the output voltage from the voltage divider increases; during inhalation, when the resistance rises, the output voltage decreases accordingly.

Figure 3 presents the results of a single breath exhaled through the mouth. As shown, a single breath is clearly detectable. The duration of the breath in both cases (a and b) is approximately 5 seconds. However, there are noticeable differences in signal amplitude between the two, as well as in the absolute slope of the signal before and after the peak. These variations may be attributed to differences in breathing intensity or individual exhalation patterns.

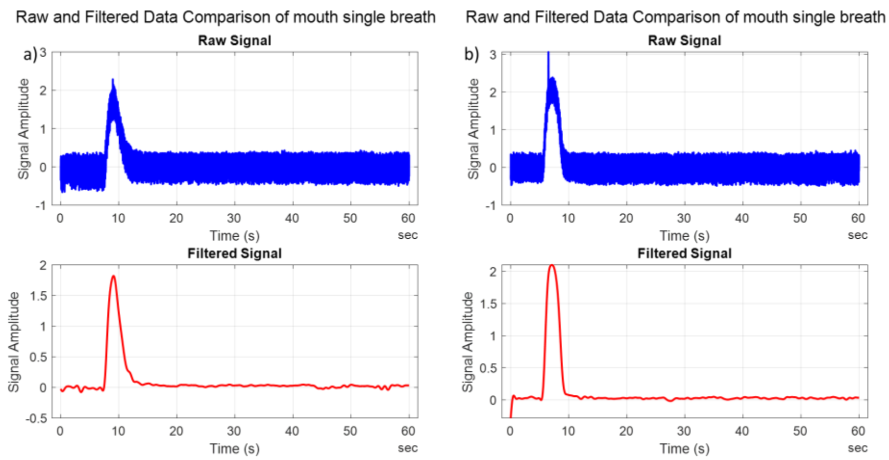


*Figure 3. Mouth single breathing a) low intensity, b) high intensity*

Moreover, the sensor was also able to detect nasal breathing patterns, as shown in Figure 4. Similar observations can be made for both parts (a) and (b) of the figure. The duration of each breath is approximately 5 seconds, which is consistent and expected, since the same person performed the test. This demonstrates the sensor's ability to reliably capture different breathing modes with clear and repeatable signals.

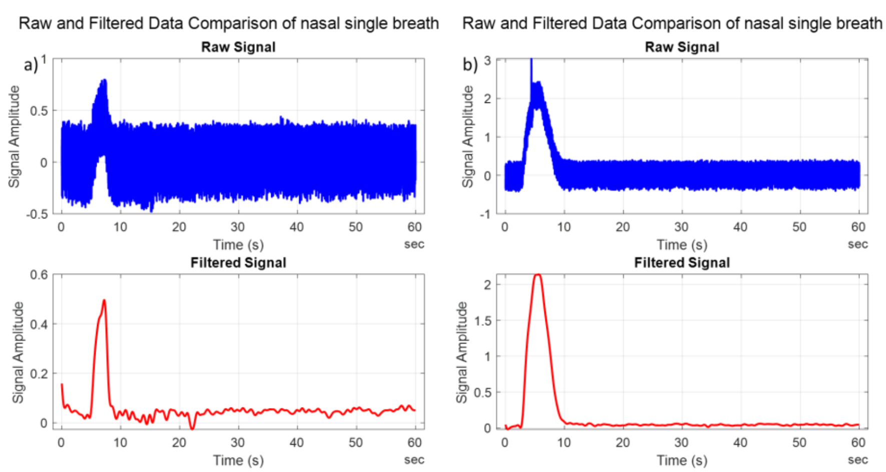


*Figure 4. Nasal single breathing a) low intensity, b) high intensity*

The main difference between parts (a) and (b) in Figures 3 and 4 is the intensity of airflow during breathing. In both figures, the exhalation and inhalation in part (a) were performed with lower intensity compared to part (b). In all cases, an effort was made to keep the breathing effort consistent to ensure a fair comparison. For mouth breathing (Figure 3), the peak amplitudes of the signals in parts (a) and (b) are quite similar, indicating that the sensor responds reliably even with slight variations in airflow. However, in nasal breathing (Figure 4), the peak amplitude in part (b) is almost five times greater than in part (a), highlighting a much stronger dependence on airflow intensity for nasal breathing. This suggests that nasal airflow produces subtler humidity changes near the sensor, making it more sensitive to variations in breathing force.

To explore more complex breathing patterns, the sensor's response was recorded over one minute of continuous mouth breathing. Figure 5 shows two normal (slow) breathing patterns with different airflow intensities, part (a) represents lower airflow, while part (b) shows higher airflow. Interestingly, the higher airflow in Figure 5b causes the sensor signal to gradually drift from its original baseline to a higher level. A possible explanation for this phenomenon is that intense exhalation not only leads to stronger, but also deeper adsorption of water molecules into the porous structure of graphene oxide. These molecules may become trapped within the material, and releasing them requires more energy, such as a stronger inhalation [7]. This dynamic highlights how airflow intensity can influence both the immediate response and baseline stability of the sensor.

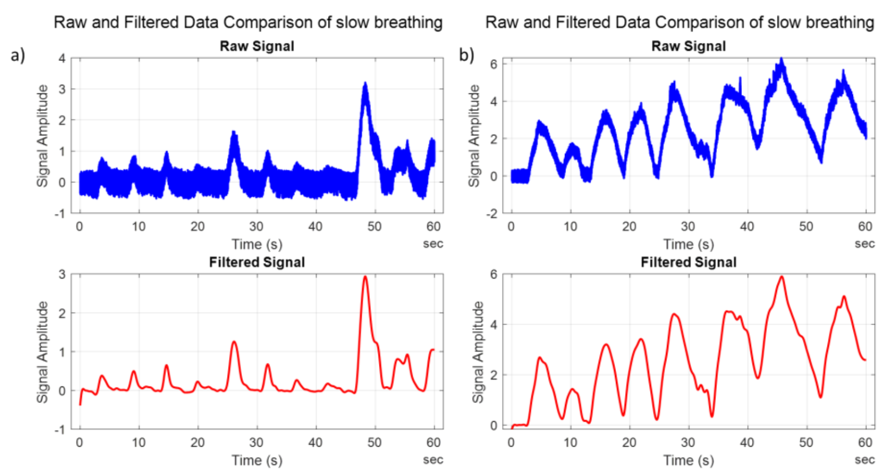


*Figure 5 normal breathing a) low intensity, b) high intensity*

To test this hypothesis, two fast breathing patterns were evaluated, and the results are shown in Figure 6. In Figure 6a, the frequency of breathing was increased while maintaining moderate intensity. In contrast, Figure 6b starts with a strong, high-intensity exhalation followed by rapid breathing. As seen in the results, the baseline in Figure 6b shifts to a higher level and the signal continues to oscillate around this new baseline. This supports the idea that high-intensity exhalation may drive moisture deeper into the porous graphene oxide structure, resulting in a lasting elevation of the baseline until the trapped water is released.

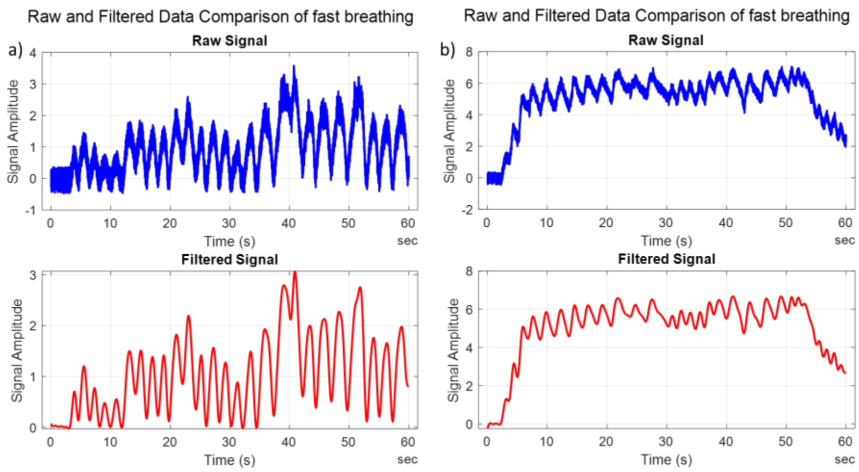


*Figure 6 fast breathing a) without and b) with high intensity exhale from start*

## Conclusion:

In this study, a graphene oxide-based resistive humidity sensor was successfully developed and tested for real-time, non-contact monitoring of human breathing. The sensor responded clearly to both exhalation and inhalation across a range of breathing patterns. The simple voltage divider readout and amplification setup enabled accurate voltage signal acquisition with minimal power consumption. Experiments revealed that while the sensor effectively tracks breathing, airflow intensity can influence the signal baseline due to moisture trapping in the porous structure of graphene oxide. This insight is crucial for future applications, particularly in wearable or continuous monitoring systems, where consistent baseline and response stability are essential. Future work should focus on optimizing sensor design to minimize drift and exploring compensation techniques for airflow variations.

### References:

[1] B. A. Kuzubasoglu, "Recent studies on the humidity sensor: A mini review," ACS Appl. Electron. Mater., vol. 4, no. 10, pp. 4797–4807, 2022.

[2] Z. Duan, Y. Jiang, and H. Tai, "Recent advances in humidity sensors for human body related humidity detection," J. Mater. Chem. C*, vol. 9, no. 42, pp. 14963–14980, 2021.

[3] H. C. Ates et al., "End-to-end design of wearable sensors," Nat. Rev. Mater., vol. 7, no. 11, pp. 887–907, 2022.

[4] M. Sajid et al, "Progress and future of relative humidity sensors: a review from materials perspective," Bull. Mater. Sci.*, vol. 45, no. 4, p. 238, 2022.

[5] D. Lei et al., "Self-powered graphene oxide humidity sensor based on potentiometric humidity transduction mechanism," Adv. Funct. Mater., vol. 32, no. 10, p. 2107330, 2022.

[6] R. Kalidoss, V. J. Surya, and Y. Sivalingam, "Recent progress in graphene derivatives/metal oxides binary nanocomposites based chemi-resistive sensors for disease diagnosis by breath analysis," Curr. Anal. Chem., vol. 18, no. 5, pp. 563–576, 2022.

[7] A. J. Carr, S. E. Lee, and A. Uysal, "Ion and water adsorption to graphene and graphene oxide surfaces," Nanoscale, vol. 15, no. 35, pp. 14319–14337, 2023.

**Appendix**: Matlab code for filtering data:

```
% MATLAB script to load, filter, and plot timetable data with 1000 Hz sampling

% Load data
load('fast2.mat'); % Replace with your file name

% Identify the timetable variable
vars = who; % Get all variables in workspace
timetable_var = [];
for i = 1:length(vars)
    if istimetable(eval(vars{i}))
        timetable_var = vars{i};
        break;
    end
end

if isempty(timetable_var)
    error('No timetable found in the loaded .mat file.');
end

% Access the timetable
TT = eval(timetable_var);

% Extract time and signal
try
    time = TT.Properties.RowTimes; % Try to get row times
    signal = TT{:,1}; % Assume first variable is signal
catch
    % If RowTimes fails, assume first column is time, second is signal
    disp('RowTimes not found. Assuming first column is time, second is signal.');
    time = TT{:,1}; % First column as time (numeric or datetime)
    signal = TT{:,2}; % Second column as signal
end

% Convert time to seconds if it's datetime
if isdatetime(time)
    time = seconds(time - time(1)); % Relative time in seconds
end

% Sampling frequency
fs = 1000; % Sampling rate (1000 Hz, as provided)

% Design a low-pass Butterworth filter
fc = 1; % Cutoff frequency (50 Hz, adjust as needed)
order = 4; % Filter order
```

```matlab
[b, a] = butter(order, fc/(fs/2), 'low'); % Butterworth filter coefficients

% Apply filter to signal
filtered_signal = filtfilt(b, a, signal); % Zero-phase filtering

% Create plots
figure('Position', [100, 100, 800, 600]);

% Plot raw data
subplot(2,1,1);
plot(time, signal, 'b-', 'LineWidth', 1.5);
title('Raw Signal');
xlabel('Time (s)');
ylabel('Signal Amplitude');
grid on;

% Plot filtered data
subplot(2,1,2);
plot(time, filtered_signal, 'r-', 'LineWidth', 1.5);
title('Filtered Signal');
xlabel('Time (s)');
ylabel('Signal Amplitude');
grid on;

% Adjust layout
sgtitle('Raw and Filtered Data Comparison of fast breathing');
```