An Easy-to-Array Respiration Sensor Based on Coupled Static Current Field

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Abstract. Respiration detection is an important detection method in the diagnosis and treatment of diseases in the future. A sensor device with respiratory dynamic characteristics sensing and chemical composition sensing will provide more information for respiration analysis. In this paper, a gas sensor based on a coupled electrostatic field and flow field is proposed. The sensor uses flow field drift and electrostatic field polarization and migration to achieve selective separation and detection of charged droplets. This sensing principle ensures that the device has a faster recovery time than a device based on the adsorption, and is more suitable for the needs of respiration detection.

Keywords: Respiration detection, Respiration sensor, device array.

1. Introduction

As a non-invasive detection method, respiration detection has attracted much attention. In addition to nitrogen, oxygen, carbon dioxide and water vapor, the exhaled breath of the human also carries various volatile organic compounds (VOCs). When the relationship between certain diseases and certain biomarker is still unclear, it will naturally require respiratory sensing devices to have higher selectivity. At present, the gas sensing technologies with high selectivity mainly include: chromatography/mass spectrometry technology and electronic nose technology [1-3]. A number of studies have used chromatography/mass spectrometry to detect breath samples or expiratory condensate to explore the possibility of early diagnosis of different types of cancer patients by respiration detection [4, 5]. Similar to this, electronic nose technology has been applied to respiration detection [6]. In order to improve the cross-specificity between cells, sensors based on different principle or different functional materials often are chose as the array cells. But high gas sensitivity makes the device preparation process very cumbersome.

This paper proposes a gas sensor array based on a coupled electrostatic field and flow field, which uses electrostatic field polarization, migration and flow field drift to achieve selective separation of different particles and generate response current. The sensor can use the difference of the flow channel to generate a differentiat physical field on each unit, thereby generating different response values and improving the cross specificity of each unit.
2. Sensor principle and structure

We have proposed a new breath sensor. The device adopts a three-electrode structure with a small distance between the upper and middle electrodes. The atomized droplets are highly polarized with a higher electric field strength and gas molecules are ionized. The middle and lower electrodes have a large distance between them and are used to collect charges. As shown in Figure 1, it is a four-layer PCB. The upper surface is covered with copper as the upper electrode, the second layer is covered with copper as the middle electrode, and the third layer is covered with S-shaped copper as the heating, and the lower surface is covered with copper as the lower electrode. Use mechanical drilling to drill a double columns of 8 rows of through holes in the center of the board. The laminated board is used as a dielectric layer between the upper, middle and lower electrodes. The electrode spacing is 0.1mm and 0.5825mm, respectively. The hole diameter is 0.4mm.

![Electrode structure of cell.](image)

3. Test experiment and result

3.1. Test experiment and experimental device

The evaporation of droplets and the polarization of the dielectric have a strong correlation with temperature [7]. In order to eliminate the influence of ambient temperature, S-type copper wires are laid in the device as a heating layer. Figure 2(A) shows the relationship curve between the temperature control voltage and the temperature. The relationship between core temperature and voltage is exponential growth, and the temperature increase of the device with carbon nanotubes is significantly faster than that of the device without the growth of nanomaterials. Figure 2(B) shows that the temperature distribution of a single-row cell on the surface of the device without growing nanomaterials under 1V, 1.75V, and 2.5V temperature-controlled voltages.

Among them, the abscissa of the curve is the distribution indicated by the red line in Figure 2(C). Under the voltage of 1.75V, the core temperature is 35.95°C, and the average temperature of the 6 valley bottoms is 34.90°C. Under 1V voltage, the core temperature is 27.14°C, and the average temperature of the 6 valley bottoms is 26.62°C. Therefore, in the temperature range from room temperature to 40°C, it is reasonable to use the center temperature of the device to represent the temperature of each unit of the device.

The gas testing device is shown in Figure (D) (E), using a pump to pump out the background gas or target gas and send it out through the sensor. The stable value of the gas flow measured by the flowmeter is 1.5L/min, the inner diameter of the gas pipe is 4mm, and the converted flow velocity is 1.989m/s. In the experiment, two metameters (Keysight B2900A precision source meter from Keysight, and KEITHLEY 2400 source meter from Hanley Technology) were used to provide voltage and measure current, single-stage constant voltage output mode, and the output voltage was 100V. Use
DC power supply (MAISHENG DC POWER SUPPLY MT-152D) as the temperature control voltage source.

![Figure 2](image)

**Figure 2.** (A) Curve of temperature control voltage and stable temperature of device center. (B) Temperature distribution on the surface of the device at 1V, 1.75V, and 2.5V. (C) Sensor substrate and detection line of temperature distribution curve. (D) Schematic diagram of gas testing device. (E) The test device. (F) Teflon packaged devices which could be installed in the test system. (G) Sensor substrate.

### 3.2. Test result

The temperature and humidity of the test environment are 21.1~25.1℃, 39.8%~51.2%. Inject VOCs such as ethanol, methanol, acetone, etc. into an air bag filled with a specific volume of air, and wait for the organic matter to completely volatilize. Before each test, purge the pipeline and sensor with background gas for 100s, and then pass in gas containing volatile organic compounds to test the response of the device to such gas. The calculated sample concentrations are listed in Table 1. The response current dropped sharply in about 20s at the beginning of the test. This may be because the device is essentially a capacitor structure, and it takes a period of time for the capacitor to discharge when a voltage is applied. The current value at the starting point is 6.34 times the reference current, as shown in Figure 3(B). Figures 3(A) and (C) are the response curves of the sensor to the four VOCs and ammonia. The sensor has a negative response to the four VOCs and the response values are 0.56, 0.488, 0.644, and 0.6 respectively. It has a positive response to ammonia (3.25).

Next, we use this device to detect human breath. In a quiet and non-interference indoor environment, we specify the distance between the position of the subject’s chin and the device to limit the error of the distance between the nose and the device in each experiment. The subjects took a uniform sitting posture throughout the whole process, and took samples for 1 minute with natural mouth breathing. Figure 5 shows the response current curve of the same unit in a short period of time using two metimeters for a designated subject. Figures 4 (A), (B) and (C) are respectively when the core temperature is 28℃, 32.2℃, and 37.2℃ Curve. The difference in the trend of the curve at 37.2℃ may be caused by the instability of the subject's two breaths. but under this temperature condition, the current amplitudes measured by the two ammeter are basically the same. This data shows that the two metameters have basically the same response to the detection of respiratory signals under the same conditions (the same array unit and two consecutive breaths of the same subject).
Figure 3. (A) Response current curve of different VOC. (B) The complete response current curve of the sensor to methanol vapor. (C) The response current curve of the sensor to ammonia water.

Table 1. Concentration ratio of experimental samples.

| Sample     | Injection volume (mL) | Concentration (ppm) | Saturated?
|------------|-----------------------|---------------------|----------
| Methanol   | 0.1                   | 7.3x10^4            | No       
| Ethanol    | 0.1                   | 9.54x10^4           | No       
| Acetone    | 0.1                   | 3.046x10^4          | No       
| Isopropanol| 0.1                   | 7.305x10^4          | No       
| Ammonia    | 0.1                   | 1.4539x10^4         | No       

Figure 4(D) uses the same element meter to measure the ambient gas response current of the same unit under different device temperatures. The device temperature is controlled by the temperature control voltage. The temperature control voltages of 2V, 1.8V, 1.5V, 1V, 0.8V correspond to the core temperature of 40.2°C, 38.4°C, 32.5°C, 26.8°C, and 26.1°C respectively. It can be observed that the higher the core temperature, the greater the baseline current of the device. The baseline current comes from the ionization current generated by the ionization of the ambient gas in the electric field, and the polarization current of the ambient gas. The higher the temperature, the higher the degree of gas ionization. Figure (A) (B) (C) also reflects another information, that is, as the temperature increases, the ratio of the response current to the baseline current is decreasing. The laboratory environment temperature is 23.3°C and humidity is 50.8%. Although the center temperature of the device controlled in the experiment is less than or equal to the normal body temperature of the human body, the expiratory airflow will quickly drop in temperature when it comes into contact with the air, and when the airflow diffuses on the device, it has become a relatively low temperature. The decrease in the ratio of the response current to the baseline current in Figures (A), (B), and (C) may be that the temperature rise increases the evaporation rate of the droplets contacting the device. The Na⁺ and K⁺ plasmas in the droplets are in the evaporation process. Neutralization reduces the net charge density in the field, resulting in a decrease in polarization current.
Figure 4. (A) (B) (C) The core temperature is 28°C, 32.2°C, 37.3°C respectively, the same unit of the device, the same test subject, the data curve of two kinds of metimeter measurement. (D) The environmental response current curve of the same unit.

4. Conclusions
In this thesis, a respiration sensor that can detect the dynamic characteristics of human respiration is designed and prepared based on the coupled physical field of static current field. It has the characteristics of MEMES preparation and miniaturization, and the sensing principle of electric field polarization makes it have a fast recovery time, which can meet the needs of breathing detection. The test results show that the sensor is sensitive to methanol, ethanol, acetone, isopropanol, and ammonia, and has a faster recovery time. The sensor has a relatively obvious response value to the human body's expiratory airflow, and as the temperature rises, the device has a better recovery time when it is close to 37°C, which can meet the recovery time requirements of respiration detection.

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