Flexible surface acoustic wave respiration sensor for monitoring obstructive sleep apnea syndrome

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Abstract

Obstructive sleep apnea syndrome (OSAS) has received much attention in recent years due to its significant harm to human health and high morbidity. A respiration monitoring system is needed to detect OSAS, so that the patient can receive treatment timely. Wired and wireless OSAS monitoring systems have been developed, but they require wire connection and batteries to operate, and they are bulky, heavy and not user-friendly. In this paper, we propose to use flexible surface acoustic wave (SAW) microsensor to detect and monitor OSAS by measuring humidity change associated with respiration of a person. SAW sensors on rigid 128°YX $LiNbO_3$ substrate are also characterized for this application. Results show both types of the SAW sensors are suitable for OSAS monitoring with good sensitivity, repeatability and reliability, and the response/recovery times for the flexible SAW sensors iare 1.125 and 0.75 s respectively. Our work demonstrates the potential of an innovative flexible microsensor for detection and monitoring of OSAS.

Keywords: Respiration sensors, Surface acoustic wave, Obstructive sleep apnea syndrome monitoring

1. Introduction

Obstructive sleep apnea syndrome (OSAS) has received much attention in recent years due to its high morbidity and significant harm to the health of persons with OSAS. OSAS patients are on rise due to increased obesity. OSAS is one of the most common sleep disorders, and is caused by full or partial obstruction of the upper airway. It is characterized by repetitive episodes of shallow or paused breathing which lasts for more than 10 s during sleep[1,2]. OSAS can lead to hypertension, coronary disease, diabetes, cardiac ischemia, myocardial infarction, congestive heart failure, stroke and nocturnal death, etc[3,4]. Early detection of OSAS allows preventative measures to be taken timely, reducing potential death. Various respiration monitoring systems have been developed for detecting OSAS, among them, polysomnography (PSG) is a widely-used device to study sleep disorders/illnesses. For instance, Philips Respironics-Alice 5 and Nicolet-32 Channel Desktop Video PSG system are two kinds of well-used PSG systems. A PSG system typically uses multi-channels to continuously record a number of biological signals such as electromyography, electroencephalography, electrocardiography, electro-oculography, nasal airflow, blood oxygen saturation, snoring sounds, and intra-esophageal pressure[4]. However, a PSG system requires a minimum of 22 wire attachments to the patient, which are connected to a computer system for recording, storing and displaying the data. The system is large, expensive, and is not suitable for ambulatory monitoring at home[5,6]. There are some portable monitoring devices on the market which record one or several biological signals, such as Portable Pulse Oximeter, Portable Sleeping Monitoring Recorder, Electrocardiography (ECG), Chest and Abdominal Movement Detection, etc. Nevertheless, they also need wires to connect sensors to the person, which will restrict patient's activity.

Use of monitoring systems with wireless communication capability can provide continuous respiration monitoring without activity restriction and behavior modification[7,8], and has been a topic of researches and developments. Xiao et al. designed a portable noncontact heartbeat and respiration monitoring system using 5 GHz radar in 2007, and this system was simplified to have only two PCB-based antennas, a palm-size PCB radio module, a data acquisition module, and a laptop[9]. Wu et al. reported a wearable textile-based wireless respiration monitoring system based on digital respiratory inductive plethysmography in 2009[5], which must be worn around thorax or abdomen for monitoring respiration, thus it is uncomfortable to users. Kumar et al. developed a Bluetooth-based wireless sensor system which could be worn as a "Band-Aid" in 2011[10]. This system has a Bluetooth module, an antenna and a rechargeable lithium polymer battery, and it is very cumbersome. Zhou et al. developed a low power miniaturized, wearable body sensor network monitoring system to monitor electrocardiograph, respiration, pulse rate, blood oxygen, blood pressure and temperature simultaneously in 2015[8]. Sohn et al. developed a 12-lead smartphone-based ECG acquisition and monitoring system to measure the respiration rate in 2017, which was composed of an analog-to-digital converter, a microcontroller board, and a Bluetooth module[11]. Although various wireless respiration detection systems have been developed in recent years, batteries, an RF communication module, data acquisition electronics, amplifying circuits and power management module are still needed to be worn for monitoring and wireless communication. The sensor chips are cumbersome and not user-friendly, therefore there is a pressing need to simplify the sensor chips.

In this work, we propose to use surface acoustic wave (SAW) sensor as flexible respiration monitoring device for detecting OSAS. Since SAW devices are very sensitive to humidity, temperature and pressure[12–14], they can be used to detect OSAS of a sleeper by measuring these respiration associated variables. Also SAW devices are a type of passive wireless sensors, thus the proposed SAW OSAS detector could be a single SAW microsensor with an integrated planar antenna attached on the wearer, whereas the other transmitter/receiver electronics are detached from the wearer. This will give the freedom of mobility to patients as the wireless communication ranges could be a few meters for SAW sensors[15]. The proposed SAW OSAS monitoring system is schematically shown in figure 1. The electronic reader sends an interrogation signal with an operation frequency of f_0 to the SAW sensor; it returns a response signal with a frequency of f_1 containing respiration information. The principle of a wireless SAW sensor is well known[16] and will not be discussed here. Flexible SAW sensors are proposed to be used for respiration monitoring owing to their conformal nature, though SAW devices on rigid substrate can also be used for this application. The cost of SAW devices is low as they can be mass-produced by microelectronic fabrication technology, therefore the proposed system can be used widely by ordinary people. Since SAW sensor is the core of the wireless, passive respiration sensor system, therefore the focus of the work is to develop suitable flexible SAW sensors for this application, not the transmitter/receiver electronics (reader) as they are commercially available [17–19]. Here, we report our initial investigation on flexible SAW-based respiration sensors, and show its excellent potential for the passive wireless OSAS detection applications. To our knowledge, this is the first time that SAW sensors are proposed for respiration detection and monitoring.

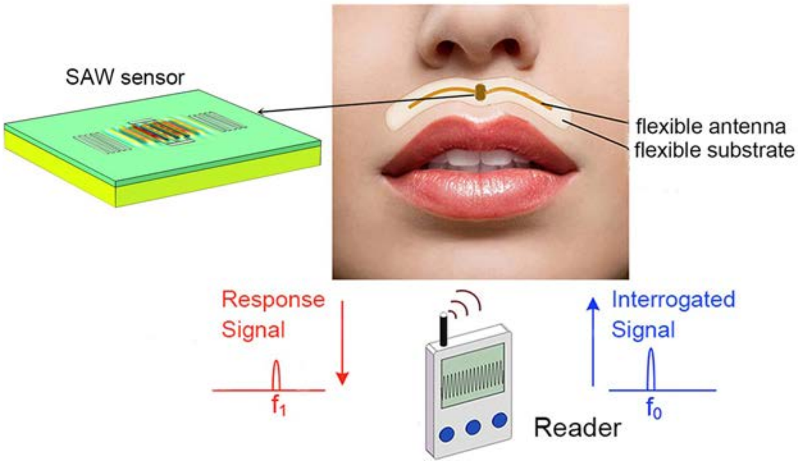

Figure 1. Schematic of the proposed wireless, passive SAW respiration sensor system.

2. Methods

2.1. Sensing principle

Exhaled air contains a high level of moisture and it is very different from that of inhaled air that has the same humidity as the surround environment. If a SAW sensor is placed on the upper lip below the nose of a person, variation in humidity of breathing air can be detected by measuring the shift of resonant frequency of the SAW sensor. By monitoring the breathing patterns through humidity change, we can detect the OSAS of the person in real-time.

2.2. SAW sensor fabrication

The flexible SAW sensors were made on ZnO piezoelectric thin films deposited on polyimide (PI) substrates using our recently developed technology[20]. $LiNbO_3$ based SAW sensors were also made for comparison, and they were designed and fabricated on 128°YX $LiNbO_3$ bulk substrates. One port resonant structure was designed for SAW sensors owing to its high quality factor (Q) and low insertion loss. Device properties of the $LiNbO_3$ SAW sensors are easy to control owing to the fixed properties of the bulk material, whereas those of the ZnO/PI SAWs depend strongly on the thickness and crystalline quality of the ZnO layer. The resonant frequency for the $LiNbO_3$ SAW was designed to be 436.4 MHz with a wavelength of 9.12 µm. The same design was used for the ZnO/PI SAW sensors, but resulted in a lower frequency of about 170 MHz due to the layered structure[21]. The device parameters of the two types of the SAW sensors are summarized in Table 1.

Table 1. Parameters of the SAW sensors.

Parameters	$LiNbO_3$ SAW tag	ZnO/PI SAW tag
Resonant frequency, f_0/MHz	436.50	170.94
Wavelength, λ / μm	9.12	9.12
Acoustic velocity, v/(m/s)	3980	1558
IDT finger pairs	50	50
IDT aperture, w/μm	100 λ	100 λ
The number of reflectors	250	250
The gap between the reflector and the IDT	1/2 λ	1/2 λ

The 128°YX $LiNbO_3$ wafers were bought from CETC Deqing Huaying Electronics Co.; while the ZnO thin films were deposited using a direct current (DC) magnetron sputtering system. The processes for preparing the flexible SAW sensors are as follows: A polyimide film (100 μm thickness, purchased from DuPont Co.) was taped on a silicon wafer for easy process. A (0002) orientation ZnO layer was then deposited on the PI layer by sputtering with a thickness of about 3 μm. The deposition conditions for the ZnO can be found from our previous publication[22]. The typical X-ray diffraction (XRD) spectrum of the ZnO film is shown in figure 2(a) with a large peak at 34.3°, corresponding to the (0002) crystal orientation. The full-width at half maximum (FWHM) of the XRD peak is about 0.173°. Grain size of the ZnO film was calculated using the Debye-Scherrer formula[23],

$$D = 0.94\lambda_x / (\beta \cos\theta) \tag{1}$$

Where λ_x is the X-ray wavelength (1.54 Å for the Cu target), β is the HWFM in radians and θ represents the Bragg angle. The calculated mean grain size is about 50 nm, similar to those previously deposited on flexible substrates[24]. Figure 2(b) is an SEM image of the cross-section of the ZnO films, showing large columnar structure of ZnO nanocrystals, perpendicular to the substrate. A microscopy image and photo of the flexible SAW sensor are shown in figure 2(c) and 2(d), respectively.

The interdigital transducers (IDTs) and reflectors were fabricated by a UV photolithography and lift-off process. The fabrication parameters for the $LiNbO_3$ SAW sensors are the same as those of the flexible SAW devices. The size of both types of the SAW sensors is 6 mm x 6 mm.

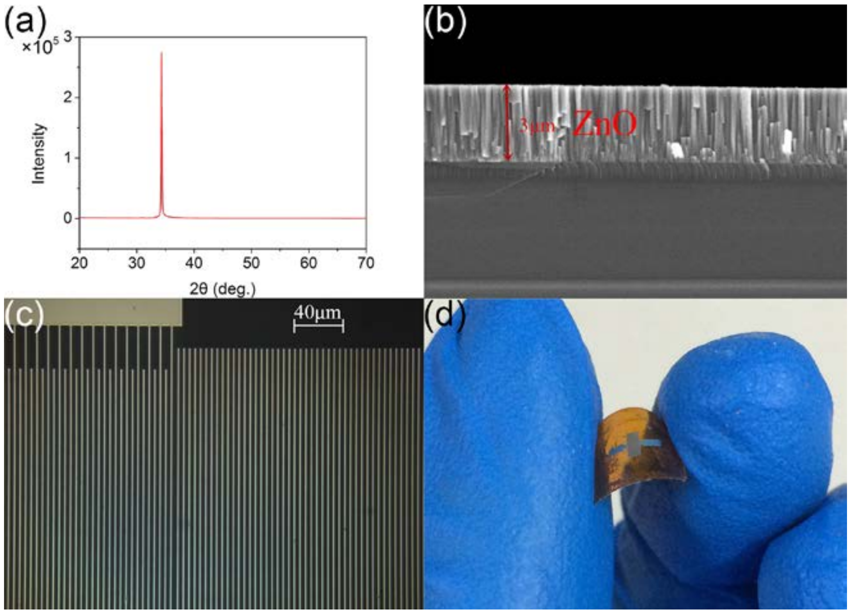

Figure 2. An XRD curve of the ZnO film deposited on a polyimide substrate (a), an SEM image of the cross section of a ZnO film (b), a microscopy image of the flexible SAW sensor (c), and a photo of the flexible SAW sensor (d).

2.3. Characterization setup

SAW sensors were characterized by a Network Analyzer (Agilent Technologies, E5071C), which was controlled by a PC. A LabVIEW (National Instruments Inc.) based program was developed on a PC to implement automated measurements to record changes in resonant frequency and return loss of SAW sensors [25]. To study the OSAS situation, we asked the volunteer to breathe discontinuously to imitate the situation of someone suffering from the OSAS. A SAW sensor was glued on a small flexible PCB, and then placed on the upper lip of the volunteer. When breathing, the expiratory air will change the temperature, humidity and pressure on surface of the SAW sensor, leading to shifts in resonant frequency and return loss. When inhaling, the resonant frequency and return loss of the SAW sensor will recover if the recovery time is shorter than the breathing period. Temperature and relative humidity of the surrounding environment were checked to be ~22 °C and ~40%RH, respectively. The humidity responses of both types of the SAW sensors were investigated, with the setup same as that used for our previous work[20].

3. Results and Discussions

3.1. SAW sensor reflection characteristics

A reflection spectrum of the $LiNbO_3$ SAW sensor is shown in figure 3(a). It has a strong resonant peak at 436.5 MHz, closed to the designed value. The reflection spectrum of the flexible ZnO/PI SAW sensor is shown in figure 3(b). There are two resonant peaks with the small one being a parasitic peak. The resonant frequency is 170.94 MHz and the return loss S_{11} is -4.7 dB for the main peak that were used for respiration monitoring.

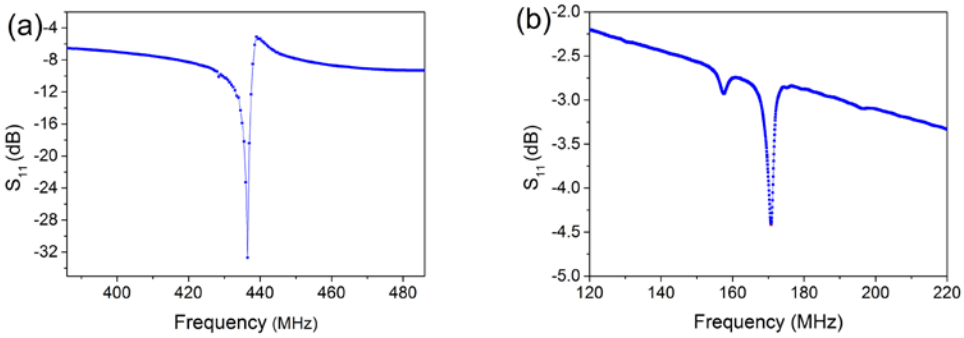

Figure 3. The reflection spectrums of the $LiNbO_3$ SAW sensor (a) and flexible ZnO/PI SAW sensor (b).

3.2. Humidity effect on reflection characteristics

Frequency shift of a SAW sensor may also be caused by variation of pressure and temperature during respiration. The SAW sensor was placed almost in parallel with the direction of breathing air, thus no pressure could be induced on the SAW sensor by expiratory air. An infrared camera (Fluke ti25, USA) was used to measure temperature of the SAW sensors during the experiments, and it was found that temperature raised about 6 degree by exhaled air.

$LiNbO_3$ and ZnO are hydrophilic materials, water molecules can be adsorbed on surface of the materials, inducing a mass loading effect on the sensors. The relative humidity of expiratory air was found to be ~93.9%RH, while that of the surrounding air was ~40%RH. Figure 4 shows the frequency change with relative humidity for both types of the SAW sensors. When the relative humidity changes from 40%RH to ~90%RH, the resonant frequency of the $LiNbO_3$ SAW sensor decreases by ~0.031 MHz, while that of the ZnO/PI SAW sensor decreases by ~0.36 MHz.

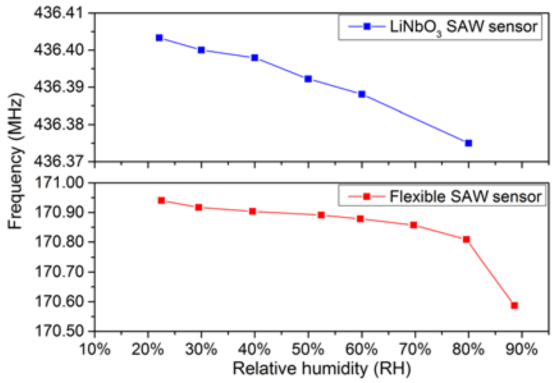

Figure 4. Resonant frequency of the two types of SAW sensors as a function of humidity.

3.3. Respiration effect on resonant frequency

Figure 5 shows the resonant frequencies of the SAW sensors on the lip under discontinuous respiration as a function of time. The resonant frequency of the two kinds of sensors shifts downwards when it receives expiratory air due to change of humidity, temperature or pressure as will be discussed later. The resonant frequency of the $LiNbO_3$ SAW sensor recovers to its original value when the volunteer inhales, implying that inhaling air has little effect on the transmission spectrum of the devices (figure 5(a)). The frequency response of the flexible SAW sensor to each breath is similar to that of the $LiNbO_3$ SAW sensor, but the overall frequency drifts downwards with time (figure 5(b)). Initially, the overall frequency drift is fast, and then slows down with time. As shown in figure 5(a), the volunteer's respiratory rate detected by the SAW sensor is about 20 min^{-1}, which is within the range of respiratory rates of a healthy adult[26]. The shift of the resonant frequency of the $LiNbO_3$ SAW sensor caused by expiratory air is about 2.7 MHz, while that for the flexible SAW sensor is about 0.1 MHz. The latter is small, but is large enough to be measured by an RF reader.

As shown in figure 5(c), the response and recovery times of the $LiNbO_3$ SAW sensor are about 1.86 s and 0.75 s respectively. The total response time for one cycle is about 2.6 s, shorter than the breathing period of ~3.0 s. Therefore, the $LiNbO_3$ SAW sensor can recover fully before next breath. For the flexible SAW sensor, the frequency cannot recover to its original value for every cycle as shown in figure 5(b). To measure the response times of the flexible SAW accurately, we conducted one cycle measurement with full recovery of the sensor with the result shown in figure 5(d). As it can be seen, the typical response time of the flexible SAW sensor is 0.58 s, but the recovery time is 3.64 s (figure 5(d)). The total time for complete recovery is more than 4.2 s, longer than the breathing period of the volunteer. As such, there is no sufficient time for the flexible SAW sensor to recover between breaths, and the resonant frequency cannot recover to its original value for each breath, but decreases continuously with respiration as moisture accumulates on surface of the flexible SAW sensor, and eventually reaches a balance. ZnO layer is polycrystalline structure with certain porosity on a flexible polyimide substrate. Both of the layers may absorb a certain amount of water molecules deep inside the films. It takes long time for water molecules to escape from inside of the films, and is believed to be responsible for the overall down drift of the resonant frequency with time. When the water absorbed by ZnO and PI layers reaches equilibrium, adsorption and desorption of water on ZnO surface of the sensor reflect the respiration.

It can be seen from figure 5(a), the volunteer held his breathes for twice. The frequency shifts of the $LiNbO_3$ SAW sensor are downwards for normal breaths, but remain unchanged during paused breathing. Figure 5(b) shows the frequency shift of the flexible SAW sensor for the discontinuous breaths. As shown, the flexible SAW sensor's frequency recovers slowly when breathing is interrupted at 38 and 78 s, respectively, and it decreases again when breathing is resumed. OSAS is characterized by repetitive episodes of shallow or paused breathing which lasts for 10 s or more during sleep. Therefore, it can be detected accurately by both types of the SAW sensors.

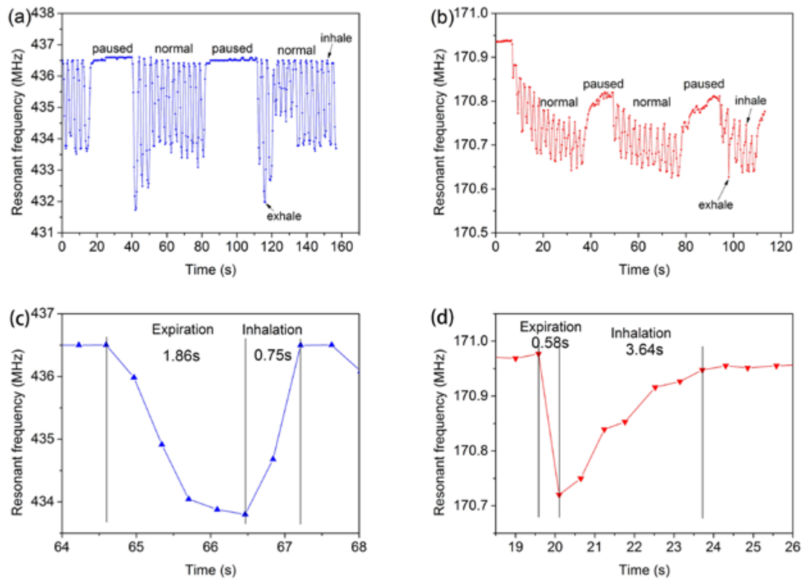

Figure 5. Resonant frequency shifts of the $LiNbO_3$ SAW sensor with breathing (a), resonant frequency shift of the flexible SAW sensor with breathing (b), resonant frequency shift for one breath of the LiNbO3 SAW sensor (c), and the flexible SAW sensor (d).

The results show that the shift of resonant frequency of the $LiNbO_3$ SAW sensors can be used for detecting the OSAS with good sensitivity, repeatability and reliability. Although the flexible SAW sensor can clearly detect periodic respiration, it may be difficult for a software to judge if it is an OSAS or not.

3.4. Respiration effect on return loss

From the results shown above, it appears that the shift of resonant frequency of the flexible SAW sensors seems not being particularly suitable as an OSAS monitoring parameter. We then investigated the relationship between return loss of the flexible SAW and respiration. Figure 6 shows variation of the return loss of the flexible SAW sensor with respiration for discontinuous breaths. The shift of loss caused by expiratory air is about 0.6±0.1 dB. The response/recovery times are about 1.125 s and 0.75 s respectively, shorter than a breathing period. The loss recovers to its original value, showing excellent repeatability and stability. As shown in figure 6, when there is no breathing, the loss remains at its original value though the amplitude varies. With the loss measurement, it is very easy for the software to judge if there are discontinuous respirations or not, demonstrating its higher accuracy, repeatability and reliability. The reason why the loss of S_{11} is more sensitive than resonant frequency as a breathing monitoring parameter is because the loss can only be influenced by change of the environment on surface of the SAW sensor.

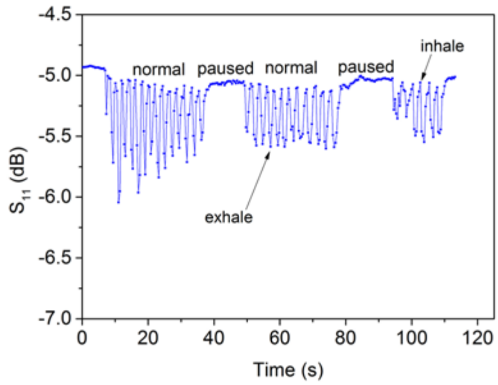

Figure 6. Loss shifts of the flexible SAW sensor with breathing.

3.5. Mechanism of SAW respiration sensors

The total shifts in resonant frequency for the $LiNbO_3$ SAW and flexible SAW sensors caused by expiratory air are about ~2.7 MHz and ~0.4 MHz respectively. The temperature of the SAW sensors was measured as shown above, and it raised only about 6 degree by exhaled air which will induce a frequency shift, but only about a tenth of the measured values, therefore the temperature influence on resonant frequency of both the SAW sensors can be ruled out.

As shown in Figure 4, when the relative humidity changes from 40%RH to ~90%RH, the resonant frequency of the $LiNbO_3$ SAW sensor decreases by ~0.031 MHz, about a tenth of the measured respiration induced shift of 2.7 MHz. The resonant frequency of the ZnO/PI SAW sensor decreases by ~0.36 MHz for the corresponding humidity change which is close to the shift of ~0.4 MHz induced by respiration. It can be concluded that the relative humidity change by expiratory air is the main cause of the observed frequency shift for the flexible SAW sensor, but it seems not to be the main one for the $LiNbO_3$ SAW sensor. Close inspection showed that condensation on surface of the $LiNbO_3$ is responsible for observed frequency shift. Expiratory air contains a high level of moisture with the temperature higher than the $LiNbO_3$ substrate, moisture liquefies on surface of the $LiNbO_3$, which induces a large mass loading effect compared to normal water molecule adsorption. Figure 7(a) and (b) are micro photos of surface of the $LiNbO_3$ SAW sensors before and after exhalation, clearly showing a high density of condensed water droplets on the surface of the SAW device, the high density of condensed water droplets on the surface would vaporize quickly when the volunteer inhales as the inhaled air flowing speed is high.

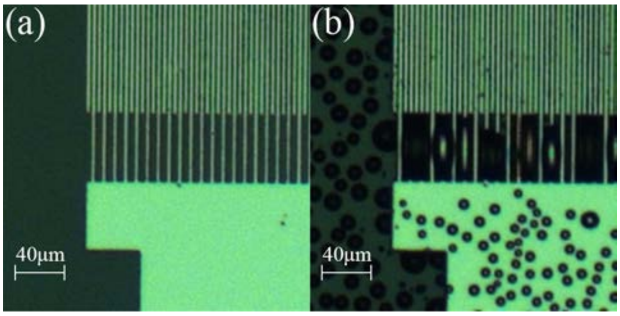

Figure 7. Micro photos of the $LiNbO_3$'s surface, before exhalation (a), after exhalation (b).

4. Conclusion

In summary, a flexible SAW sensor has been proposed for monitoring respiration that is suitable for OSAS detection and monitoring application. Two types of SAW sensors have been fabricated and assessed with one on flexible ZnO/PI and another on $LiNbO_3$ bulk substrate. The results showed that the shift of resonant frequency of the $LiNbO_3$ SAW sensors is suitable for detecting respiration and OSAS, while the shift of return loss of the flexible SAW sensors is more suitable for this application. The response/recovery times for the flexible SAW sensor are 1.125 and 0.75 s respectively. The sensitivities for the $LiNbO_3$ SAW and flexible SAW are about ~2.7 MHz/50%RH (induced by condensation on surface) and ~0.36 MHz/50%RH respectively. Both types of the SAW sensors have good sensitivity and excellent repeatability for the respiration and OSAS monitoring. The gain of the present flexible SAW sensors is low, and more efforts are needed to improve the performance of the flexible SAW sensors, which may be achieved by using impedance matching design, high quality thick ZnO film and optimized fabrication process. Additionally, the relative humidity of environment would not change abruptly, it will affect the magnitude of frequency shift a little, but the distinction between the test results of OSAS and normal breath will still be obvious. This initial work has clearly demonstrated the feasibility of a wireless passive sensor for OSAS monitoring and detection.

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