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ORIGINAL RESEARCH

Polyvinylidene fluoride film based nasal sensor to monitor human respiration pattern: An initial clinical study

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Abstract Design and development of a piezoelectric polyvinylidene fluoride (PVDF) thin film based nasal sensor to monitor human respiration pattern (RP) from each nostril simultaneously is presented in this paper. Thin film based PVDF nasal sensor is designed in a cantilever beam configuration. Two cantilevers are mounted on a spectacle frame in such a way that the air flow from each nostril impinges on this sensor causing bending of the cantilever beams. Voltage signal produced due to air flow induced dynamic piezoelectric effect produce a respective RP. A group of 23 healthy awake human subjects are studied. The RP in terms of respiratory rate (RR) and Respiratory air-flow changes/alterations obtained from the developed PVDF nasal sensor are compared with RP obtained from respiratory inductance plethysmograph (RIP) device. The mean RR of the developed nasal sensor (19.65 $\pm$ 4.1) and the RIP (19.57 $\pm$ 4.1) are found to be almost same (difference not significant, $p > 0.05$) with the correlation coefficient 0.96, $p < 0.0001$. It was observed that any change/alterations in the pattern of RIP is followed by same amount of change/alterations in the pattern of PVDF nasal sensor with $k = 0.815$ indicating strong agreement between the PVDF nasal sensor and RIP respiratory air-flow pattern. The developed sensor is simple in design, non-invasive, patient friendly and hence shows promising routine clinical usage. The preliminary result shows that this new method can have various applications in respiratory monitoring and diagnosis.

Keywords PVDF thin film · Nasal sensor · Cantilever · Human respiration pattern · Respiratory inductance plethysmograph

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1 Introduction

Sensors for biomedical signal monitoring with the capability of quantitative diagnosis is a rapidly growing field of research. Especially, the use of piezoelectric sensors for applications in biomedical signal monitoring has been well documented [1]. Continuous monitoring of vital signs is important for the diagnosis of different diseases or conditions, for which, suitable monitoring method can assist the physicians to identify whether or not the patient is in a healthy or unhealthy state [2]. Folke et al. [3] has earlier emphasized the need of non-invasive respiratory monitoring in medical health care. Early changes detected in critically ill patients by monitoring the Respiratory Rate (RR) as a vital sign has been studied [4]. Apart from monitoring RR, changes in respiration in terms of depth/magnitude can provide important information about complete or partial nasal obstruction [5]. In addition, certain changes/alterations in the Respiratory Pattern (RP) may also provide information about sleep apnea [6].

The measurement of RR in routine clinical practice is done subjectively, based on direct observation of chest wall movement, despite the fact that it may be highly inaccurate [7, 8]. Other than subjective observations, several objective devices are also used to monitor the RR, such as impedance pneumography [9], acoustic sensing [10], fiber optic sensing [11], ECG [12], and end tidal O_2 and CO_2 measurement [13]. Impedance pneumography and ECG monitors respiration as a change in trans-thoracic impedance. An acoustic sensor and fiber optic sensor work on the principle of sound of airflow and humidity during expiration, respectively, to monitor respiration. Monitoring the O_2 and CO_2 concentrations of the exhaled air can also be helpful in assessing the respiratory activity. However, certain problems arise in clinical usage of these devices such as being extremely sensitive to patient's movements, having a slow response time, being expensive and having poor quality of respiratory tracing [14, 15]. Apart from the above RR measurement devices, Respiratory Inductance Plethysmograph (RIP) is one of the popularly used methods to monitor respiration [16]. This method enables to monitor respiration noninvasively by assessing the change in the cross-sectional areas of the thoracic and abdominal compartments using two flexible sinusoidal wires.

Kawai discovered the strong piezoelectric property of polyvinylidene fluoride (PVDF) in 1969 [17]. It is a highly non-reactive, flexible, light-weight, tough engineering plastic and pure thermoplastic flour-polymer that can be prepared in a wide variety of thickness and sizes. A PVDF based sensor has many advantages as compared to other sensors used for monitoring respiration. It is comparatively inexpensive, and because of its piezoelectric and pyroelectric properties, it does not require external power source [14]. PVDF sensor is positioned at various places like inside the oxygen mask [14, 18, 19], under the bed [20, 21] and under the chest belt [22, 23] to monitor the changes in respiration. The pyroelectric property of the PVDF is the ability to produce voltage signal proportional to the temperature differences in the nasal airflow. Dodds et al. [14], Huang et al. [18] and Kulkarni et al. [19] have used this pyroelectric property of the PVDF film to monitor respiration by placing them inside the oxygen mask. Similarly, the piezoelectric property of the PVDF is the ability to produce output voltage resulting from dynamic mechanical pressure. This piezoelectric property was used by Siivola [20] to monitor the RR by placing the PVDF film under bed. Similar measurements were carried out by Niizeki et al. [21] by placing an array of eight PVDF film under the bed to measure cardiorespiratory activity. Also, Karki and Lekkala [22] and Choi and Jiang [23] developed a wearable device consisting of a belt sensor with PVDF for measuring respiratory cycle.

The aim of our present study is to design the PVDF film based nasal sensor using its piezoelectric property to monitor the human respiration pattern of each nostril simultaneously. This is done by using two identical PVDF films in a cantilever configuration and by subjecting them to dynamic pressure due to airflow from each nostril. We have also evaluated the results of PVDF nasal sensor by comparing these with RIP technique in normal subjects. The sensor reported in this paper is simple in design, non-invasive, compact, easily portable, patient friendly, and can be used routinely to monitor respiratory pattern by clinicians.

2 Device and methods

2.1 Subjects

A group of 23 healthy awake human subjects, 15 male and 8 female (with no history of respiratory related disease and medication free) with age of 26 $\pm$ 4 (mean $\pm$ SD) years and Body Mass Index of 22 $\pm$ 2 kg/m^2 (Weight 62 $\pm$ 8 kg, Height 1.6 $\pm$ 0.08 m) (mean $\pm$ SD), were included in the study and sample data were collected at M S Ramaiah Medical College and Hospital, Bangalore, India. Ethical clearance was obtained from the institutional review board for the study protocol. Written consent was obtained from all subjects prior to their participation in the study.

2.2 Nasal sensor design

The PVDF film used in this work is a stretched, poled film with gold coating as the electrode on both sides. One end of the PVDF film with length 10 mm, width 5 mm and thickness 28 μm (optimization of dimensions explained in Sect. 2.3) is firmly adhered to a plastic base in such a way that it forms a cantilever configuration leaving the other end free for deflection. The double enameled copper wires (diameter 0.07 mm), whose ends were tinned about 3 mm length are attached on top and bottom surfaces of the PVDF thin film using aluminum conducting tape for measurement of voltage output. The voltage output is recorded using a data acquisition system. PVDF film in cantilever configuration with the leads attached on the surface (on both sides) becomes the PVDF nasal senor. The plastic base of the PVDF nasal sensor is soldered to a flexible string so that the PVDF nasal sensor can be adjusted as per requirement. Thus, the two identical PVDF nasal sensors with flexible strings are mounted on an ordinary spectacle frame as shown in Fig. 1, in such a way that these sensors are placed about 4 mm below the nostrils without disturbing the normal breathing. The pulsating air flow due to the inspired and expired air impinge on these two identical PVDF nasal sensors at the respective nostril leading to bending strain that generates voltage signal corresponding to respiration pattern of the respective nostrils.

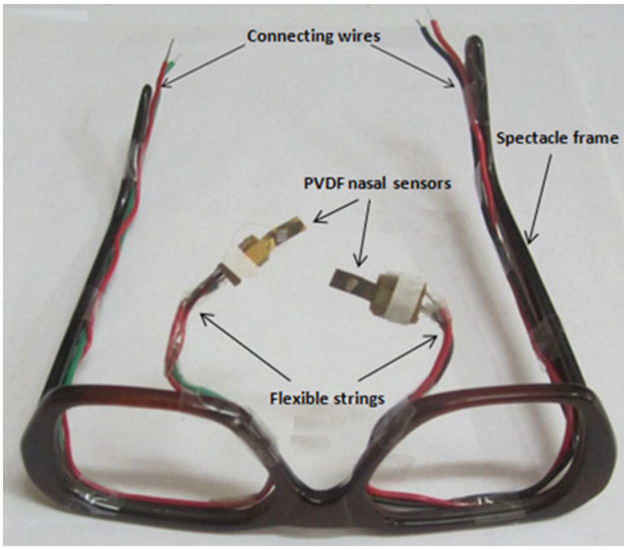

Fig. 1 PVDF nasal sensor mounted on a spectacle frame

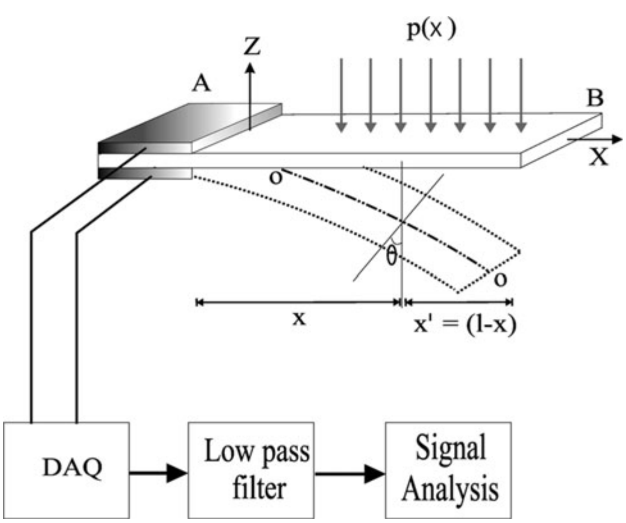

Fig. 2 Cantilever beam with aerodynamic load with sensing voltage measurement circuit

2.3 Sensing principle

In the present paper, only piezoelectric property of the PVDF film is considered for the analysis. However, PVDF is known for its pyro-electric property, and related effects due to body temperature variation etc. will be reported elsewhere. The PVDF film is free of any substrate while its one end 'A' is fixed leaving the other end 'B' to deflect like a cantilever, as shown in Fig. 2. When the cantilever beam is subjected to an aerodynamic pressure, it bends in such a way that the area above the neutral axis 'OO' is elongated and the area below neutral axis 'OO' is compressed. Since it is piezoelectric in nature, the bending strain causes a charge flow from one electrode to the other electrode placed on the two faces, giving rise to a voltage signal which was measured through a set-up as shown in Fig. 2.

Consider a point 'u' at a distance of 'x' from the fixed end of the cantilever beam. The displacement in x-direction due to bending is given by

$$u(x,z) \cong -z\theta = -z\frac{dw(x)}{dx} \tag{1}$$

where w(x) is the transverse deflection in z-direction. The longitudinal strain ε_{xx} due to bending can be expressed in terms of deflection as

$$\varepsilon_{xx} = \frac{du}{dx} = -z\frac{d^2w(x)}{dx^2} \tag{2}$$

In the Cartesian co-ordinate system fixed at the cantilever root, the material model for a linear piezoelectric material is given by

$$\sigma_{xx} = C_{11}\varepsilon_{xx} + e_{31}E_3 \tag{3}$$

$$D_z = e_{31}\varepsilon_{xx} + \in_{33} E_3 \tag{4}$$

where σ_{xx} denotes the stress, C_{11} denotes the elastic modulus of the piezoelectric material, e_{31} denotes the piezoelectric constants, D_z denotes the electric displacement, ε_{33} denotes the dielectric permittivity of the piezoelectric material, E_3 denotes the electric field. The equilibrium expressions for the bending moment for the section shown in Fig. 2 can be written as

$$\int \sigma_{xx} bz dz + \int \frac{1}{2}\rho v^2 \alpha b x' dx' = 0 \tag{5}$$

where b is the width of the cantilever beam, $p(z) = \frac{1}{2}\rho v^2 \alpha$ is the aerodynamic force on the cantilever beam with ρ as the air density (1.184 kg/m^3), v as the air velocity in m/sec, α is the pressure coefficient as function of the shape and the nature of air flow. By substituting Eq (3) in Eq (5) and solving the resultant equation gives the deflection equation of the cantilever beam as

$$w(x) = \frac{\rho v^2 \alpha b}{48C_{11}I}\left[x^4 + 6l^2x^2 - 4lx^3\right] \tag{6}$$

where l is the length of the cantilever beam, I = bh^3/12 denotes the second area moment of the beam cross section with thickness h and width b. Now the sensor (PVDF cantilever) is exposed to aerodynamic pressure field which generates a charge q in response, which is measured as sensor voltage ΔV and is expressed as

$$\Delta V = \frac{q}{C_p} \tag{7}$$

where C_p is the effective capacitance of the piezoelectric cantilever. The charge collected by the electrode on the surface depends on the component of the electrode area normal to the electric displacement, that is,

$$q = \iint_A (e_{31}\varepsilon_{xx} + \in_{33} E_3)dA \tag{8}$$

The deflection of the nasal sensor produces the voltage due to its piezoelectric property. While breathing, the aerodynamic pressure exerted on the sensor causes the deflection producing a corresponding voltage. The voltage signal is the potential difference between the top electrode and the bottom electrode and is given by

$$\Delta V = V(h/2) - V(-h/2) \tag{9}$$

with the help of Eqs. (7) and (8) substituting in Eq. (9) we get an expression for the total differential voltage output from the sensor as

$$\Delta V = \frac{\rho v^2 l^3 e_{31} \alpha b}{C_p C_{11} h^2} \tag{10}$$

2.4 Optimization of the PVDF cantilever dimension

Since PVDF film is used in a cantilever configuration, in order to optimize its dimensions, bending deflection at the tip of a simple cantilever is considered. This is given by Eq. (6). The deflection at $x = l$ can be expressed as

$$w(l) = \frac{\rho v^2 \alpha b l^4}{16 C_{11} I} \tag{11}$$

Also, a uniform aerodynamic pressure is assumed to be exerted by each nostril on the PVDF cantilever film. Optimization of the PVDF cantilever dimension is done using Eq. (11). As the length of the cantilever increases, the deflection also increases as can be seen in Fig. 3a. If we consider length beyond 10 mm, the two cantilevers may touch each other and may not be free for deflection. The cantilever width does not have any significant effect in its deflection as it is a constant function of deflection. So the width is chosen as 5 mm which is less than the length. As the PVDF cantilever thickness increases, the deflection decreases as can be seen in Fig. 3b. Though for 9 μm thickness the deflection is more, but it is too thin and is very sensitive to small vibration, hence the next available thickness of 28 μm was chosen as PVDF cantilever dimension. Based on the Fig. 3a, b, the length, the thickness and the width of the PVDF cantilever are optimized as 10 mm, 28 μm and 5 mm respectively.

2.5 Experimental setup

Figure 4 shows the block diagram of the experimental setup. In order to evaluate the performance of the developed PVDF nasal sensor, it was compared with the "gold standard" RIP to monitor the respiratory movement. Each

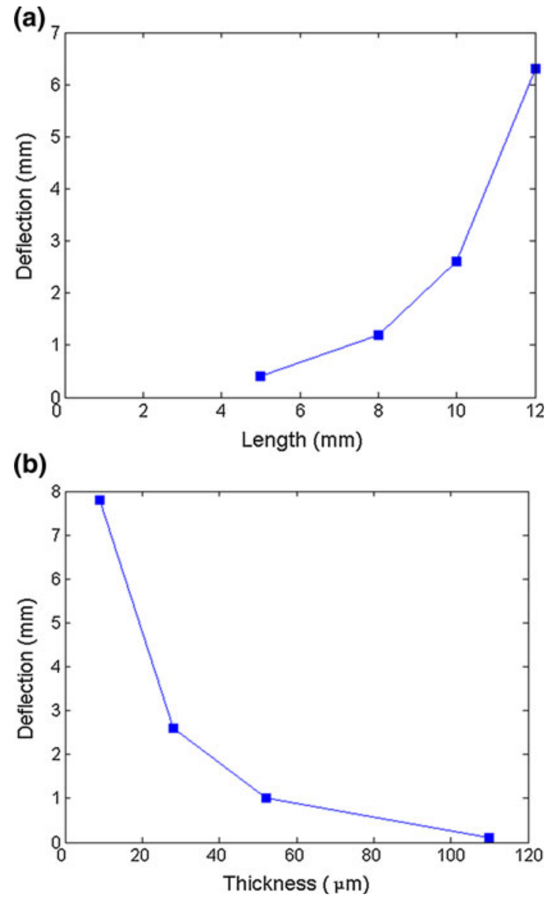

Fig. 3 PVDF cantilever tip deflection under uniform aerodynamic pressure (**a**) deflection versus length for constant width of 5 mm and thickness of 28 μm (**b**) deflection versus thickness for constant length of 10 mm and width of 5 mm

subject was made to sit in a relaxed position in a room. After wearing the spectacle frame mounted with PVDF nasal sensor and RIP (by wrapping the inductance bands around the chest and abdomen) by the subject, initial 30 s data was discarded in order to reduce the effects caused due to wearing of the sensors on the subject's face and body. Once the signal became artifact free, the RP for each subject was simultaneously recorded using the PVDF nasal sensor and RIP for 120 s (i.e. 2 min) and stored in two separate computers (one for PVDF nasal sensor and the other for RIP) with a sampling rate of 1.25 and 1.02 kS/s, respectively. While recording RP, subjects were asked to do a quiet breathing with their mouth closed. The quiet breathing consisted of about 30 s (5–6 times) of normal breathing, then about 30 s (4–5 times) of voluntary increased tidal breathing, then about 30 s (3–4 times) of voluntary decreased tidal breathing and again about 30 s (5–6 times) of normal breathing. The subjects were asked to breathe in this way intentionally, for the purpose of comparing the change in the respiratory air flow pattern between the PVDF nasal sensor and the RIP.

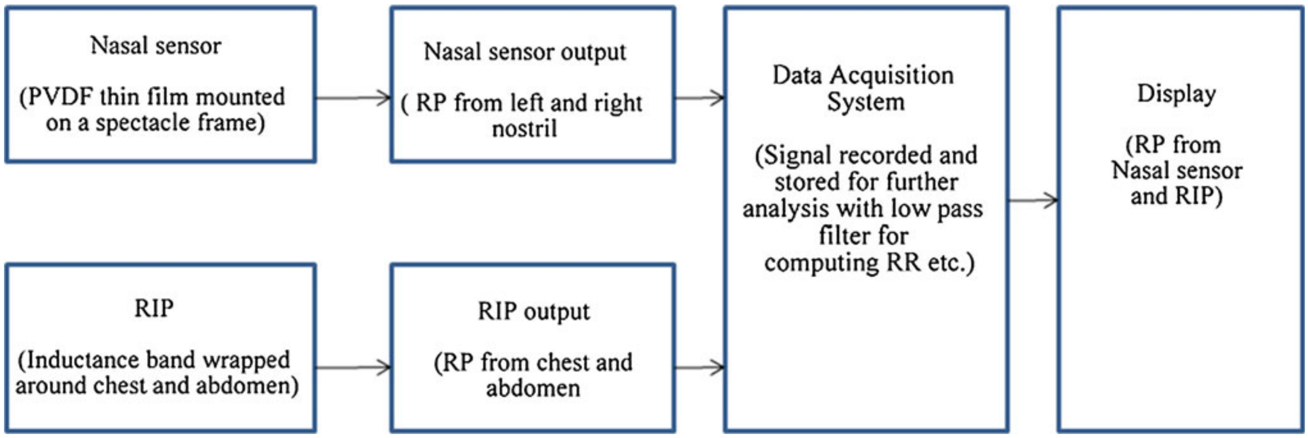

Fig. 4 Block diagram of a complete PVDF nasal sensor system with provision for comparison with gold standard RIP device

2.6 Analysis of respiratory signal

The respiratory signal analysis was done for 60 s data out of 120 s for easy calculations. The RP captured by the PVDF nasal sensor consists of respiratory parameters like RR, respiratory air flow pattern and respiration magnitude/depth. The stored files of PVDF nasal sensor and the RIP outputs (raw data) were processed through low pass filter with the cut-off frequency of 20 Hz using LabVIEW software. Further, a computer algorithm was developed using MATLAB software for computing RR, depth/amplitude and changes in the respiratory air flow pattern. The RR is the number of respiration cycle per minute. The respiration magnitude/depth is the peak-to-peak amplitude of the respiration cycle, which corresponds to the magnitude of air-flow from each nostril. The respiratory air flow pattern in terms of voltage signal gives a pictorial representation of the flow going up or down i.e., the waveform changes (alterations). The RP captured by the RIP consists of respiratory parameters like respiratory rate (RR) and air flow pattern. But it does not give the magnitude/depth of the respiration from the nostril because the RP captured by the RIP is due to change in volume of chest and abdomen and not due to nasal breathing. Therefore a direct quantitative correlation is difficult for nasal magnitude/depth respiration for RIP.

2.7 Statistical analysis

Statistical analysis was carried out using SPSS 18 (Statistical Package for Social Sciences version 18.0 for windows) and descriptive statistics such as mean, Standard Deviation (SD) were computed. In order to test for the differences in the mean values between the nasal sensor and the RIP, paired t test was employed and statistical significance was accepted at $p < 0.05$. Further, for the purpose of understanding the relationship between two quantitatively measured variables with nasal sensor and RIP, the bi-variate correlation coefficient was computed. A Bland–Altman plot was constructed to compare both systems (PVDF nasal sensor and RIP) for measuring RR [24]. A kappa statistics (κ) was used for agreement analysis for the change (alterations) in RP obtained between the PVDF nasal sensor and RIP device [25]. An event was defined in the 60 s long RP of the PVDF nasal sensor and the RIP, when there was any changes/alterations i.e., increase/decrease in the tidal airflow lasting >5 s. A 2 × 2 table was constructed for the number of events identified by the nasal sensor and the RIP device, the nasal sensor but not the RIP device, the RIP device but not the nasal sensor, and neither the nasal sensor nor the RIP device.

3 Results

The RP of 23 (with mean age of 26 ± 4 years) healthy subjects were captured using PVDF nasal sensor and RIP. Among 23 healthy subjects, 15 were male subjects with mean age of 28 ± 4 years and 8 female subjects with the mean age of 24 ± 4 years. The sample tracing of the RP's of 2 healthy subjects obtained from the developed PVDF nasal sensor (from right nostril and left nostril) and the RIP device which is used as gold standard are shown in Fig. 5a, b. In this example of RP, the curve above the zero line denotes the expiration phase while the curve below the zero line denotes the inspiration phase. The voltage signal has been normalized in both the methods for comparison purpose. The decreased tidal respiration (lasting >5 s), increased tidal respiration (lasting >5 s) and normal tidal respiration are marked as A, B and C respectively, in the RP of both the methods. Similarly, the alterations in the tidal respiration and no tidal respiration (lasting >5 s) are marked as D and E respectively are shown in Fig. 5b. From Table 1, the computed value of level of kappa agreement (κ) between the two

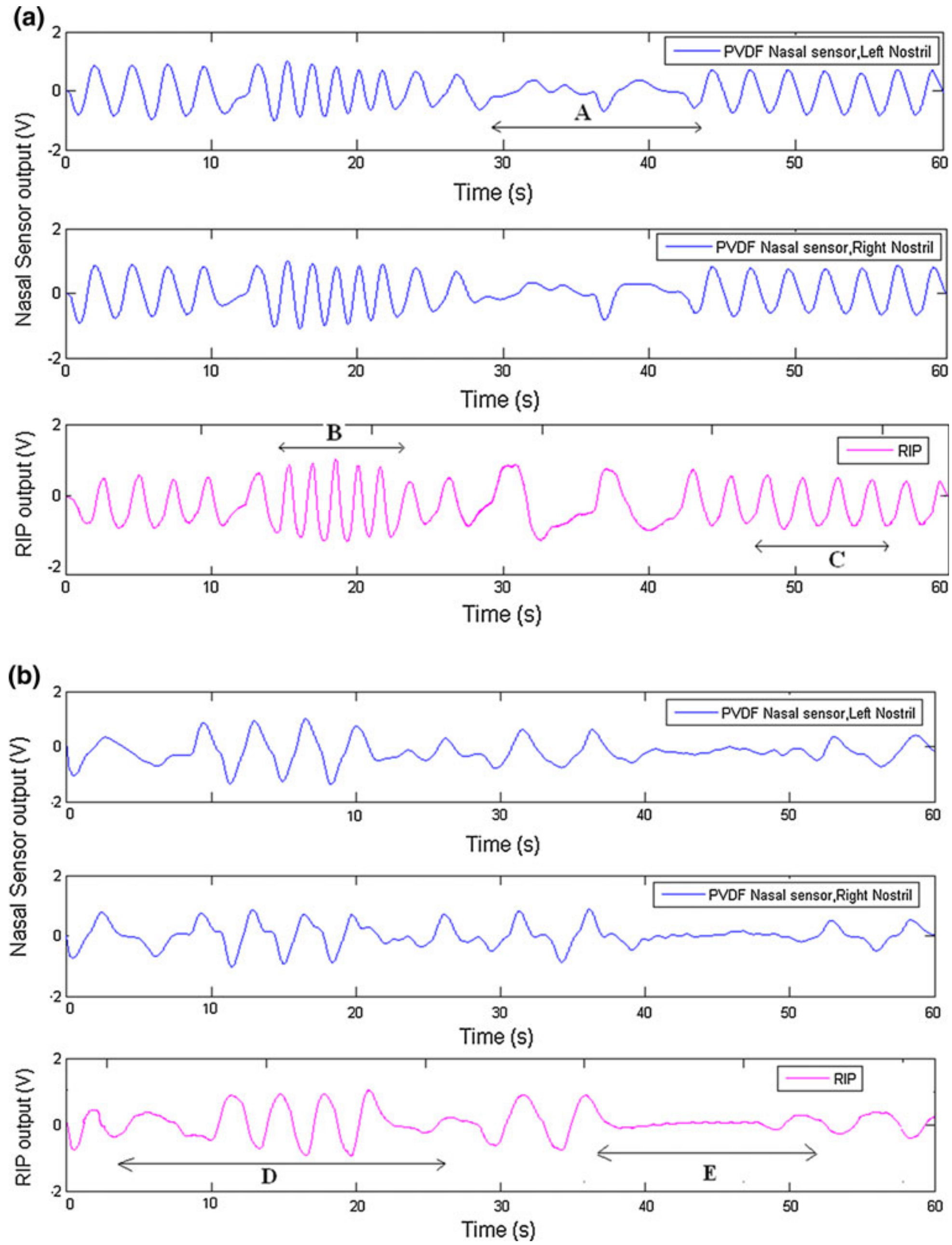

Fig. 5 An example of a RP traced from the PVDF nasal sensor (*right nostril* and *left nostril*) and the RIP device. A 60 s output is shown with alterations (>5 s) in the respiration for (**a**) subject 1 and (**b**) subject 2 (*A–E* indicates various features in the RP: *A* decreased tidal respiration, *B* increased tidal respiration, *C* normal tidal respiration, *D* alterations in tidal respiration, *E* no tidal respiration)

Table 1 Level of agreement between the PVDF nasal sensor and the RIP

PVDF nasal sensor	Respiratory impedance plethysmograph	
	Yes	No
Yes	48	2
No	3	16

Table 2 Average RR measured using PVDF nasal sensor and RIP, for male and female with their level of significance

Gender	Avg. RR measured using PVDF nasal sensor	Avg. RR measured using RIP	Significance level (p)
Male (n = 15)	18.93 ± 3.71	18.83 ± 3.58	0.958
Female (n = 8)	21 ± 4.92	20.86 ± 5.08	0.959

sensors: PVDF nasal sensor and RIP was found to be 0.815 with sensitivity of 94.11 % and specificity of 88.88 %.

As can be seen in Table 2, the RR was different for male (n = 15) and female (n = 8) subjects. There was no statistical significance between the PVDF nasal sensor and RIP for the RR measurement in male and female ($p > 0.05$) subjects. The mean (±SD) value of all subjects RRs

Table 3 Correlation between the PVDF nasal sensor and the RIP

Parameter	PVDF Nasal Sensor (NS)			RIP	Correlation co-efficient (NS vs. RIP)	Significance level
	Right NS	Left NS	Mean			
Respiration rate (in min)	19.67 ± 4.1	19.64 ± 4.1	19.65 ± 4.1	19.84 ± 4.1	0.96	$p < 0.0001$

(n = 23) of the developed PVDF nasal sensor i.e., the right nostril nasal sensor and the left nostril nasal sensor were 19.67 ± 4.1 and 19.67 ± 4.1, respectively. The mean RR of these two PVDF nasal sensors is taken for the comparison with the RR of RIP. Therefore, the mean (±SD) RR of the PVDF nasal sensor and the RIP device were 19.65 ± 4.1 and 19.57 ± 4.1, respectively. The difference in the mean RR values between the two appliances was found to be statically not significant ($p > 0.05$). As shown in Table 3, the correlation coefficient between the nasal sensor and the RIP device was calculated and found to be 0.96 with $p < 0.0001$ showing a strong relationship between the nasal sensor and the RIP device. A scatter plot is shown in Fig. 6, by taking the nasal sensor data as dependent variable and the RIP data as independent variable. The regression line was found to be y = 0.6545 + 0.9710× and these entire scatter plots were found to be almost near to the regression line, indicating a good relationship of prediction equation. Though the value

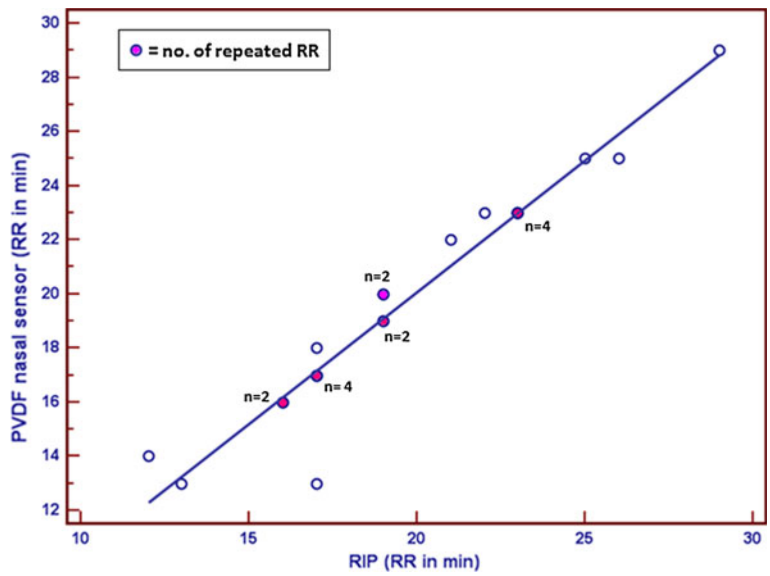

Fig. 6 A *scatter plot* showing the strong relationship between the nasal sensor signal and the RIP signal. The *solid line* is the regression line (r = 0.93, $p < 0.001$)

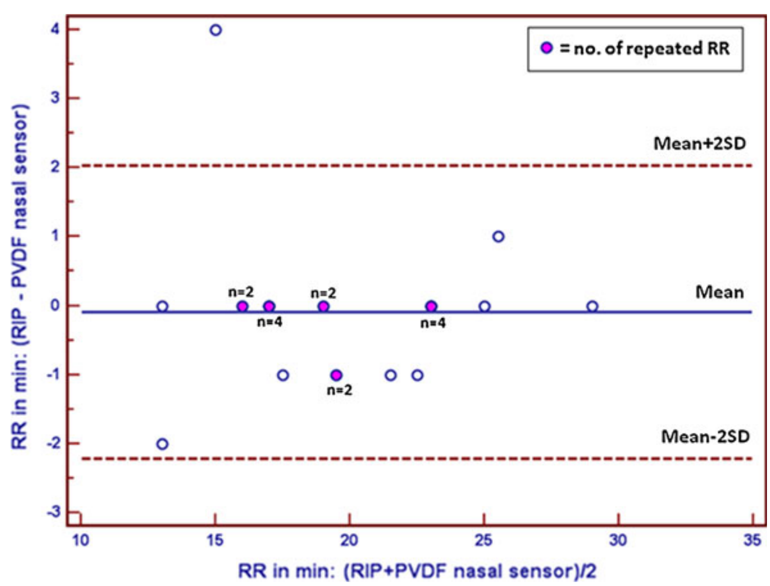

Fig. 7 Plot of the Bland–Altman agreement analysis. On the y-axis are represented the difference in the RR for each subject (RIP-Nasal sensor) plotted against the average of the two methods (RIP + Nasal sensor)/2 are represented on the x-axis

of correlation coefficient is high, a Bland–Altman plot was also plotted as shown in Fig. 7 for the mean differences (RIP -nasal sensor) versus the average value (RIP device + nasal sensor)/2 for the respiratory rate. In Figs. 6 and 7, some points (RR) were repeated more than once and hence these points were colour coded.

4 Discussion

The PVDF sensor is taken in cantilever configuration to make efficient use of its piezoelectric property. When a cantilever beam is subjected to uniform aerodynamic pressure, during expiration phase the tensile stress acts on the exposed surface of the PVDF cantilever. Voltage with positive polarity is developed, as the PVDF thin film is polarized in the z-direction (thickness direction). Similarly, during inspiration phase, a compressive stress acts on the exposed surface of the PVDF cantilever giving rise to a voltage with negative polarity. Therefore, when the PVDF nasal sensor is continuously exposed to a respiratory cycle, a corresponding RP is developed as shown in Fig. 5a, b. This pattern is in good agreement with respiration pattern traced by PZT film attached on a human belly [26]. The nasal airflow is thus monitored by measuring the variation in the voltage of the PVDF sensor, as this is exposed to the dynamic pressure which is directly related to the magnitude/depth of the flow [27]. It is important to note that, setting up of the PVDF nasal sensor in cantilever configuration is simple and it exhibits excellent dynamic response.

In the present work, the main objective was to implement the PVDF cantilever as a nasal sensor and capture/monitor the human respiratory pattern from each nostril. We have successfully evaluated the results of PVDF nasal sensor, by comparing it with RIP technique in normal subjects. The accuracy of RIP is far better than those of impedencemetry, obtained with ECG [28]. Electrodes used for ECG impedencemetry are smaller, less adhesive and are made from different materials may be prone to large artifacts during patient movements, whereas Inductance sensors are comfortable and mechanically stable [28]. Dodds et al. [14] mentioned that the respiratory curve traced by the ECG is of poor quality and the impedance changes in the skin/electrode interface. During natural breathing, RIP provides accurate measures of respiratory parameters [29, 30]. Therefore RIP has been used in our study as "Gold standard" comparison method.

We have compared the RR of the designed PVDF nasal sensor (19.65 ± 4.1) with the "gold standard" RIP (19.57 ± 4.1). Good agreement among these two methods was established using a Bland–Altman plot shown in Fig. 8. It can be seen that all the points (RR) lie within ± 2 SD, having only one point outside, which is acceptable and indicates that the new method is as accurate as the gold standard. Therefore, the new device reported in our present work can be used clinically. Generally the RR for a female will be more than that of a male because of their anatomically average lower tidal volume in comparison with

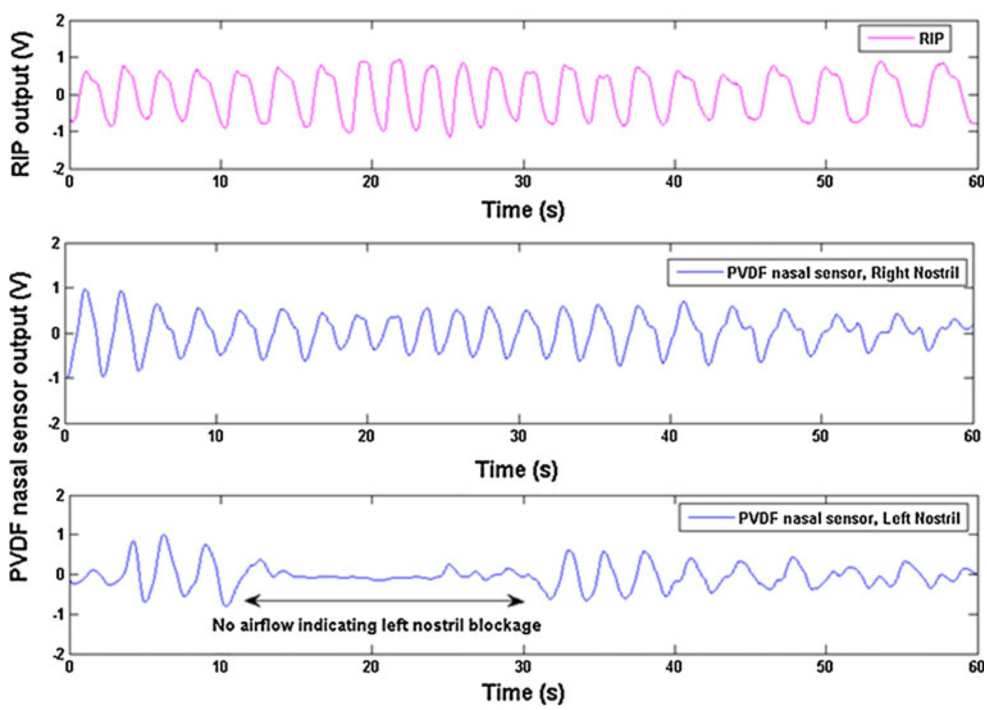

Fig. 8 An example of a RP traced from the PVDF nasal sensor (*right nostril* and *left nostril*) and RIP device in which one can observe that there is a minimal airflow >10 s from left nasal cavity as it has partial blockage

the male [31]. Therefore, the RR of the female was higher than that of the male as can be seen in Table 2. The mean RR measured by the designed PVDF nasal sensor for the healthy subjects is within the normal range as reported in the literature [32].

In our present study, all subjects were asked to breath normally for about 30 s (about 5–6 times), to breath fast (increased tidal respiration) for about 30 s (about 4–5 times) and to breath slow (decreased tidal respiration) for about 30 s (about 3–4 times) and then again about 30 s (about 5–6 times) of normal breathing in sequence. These different types of breathing are considered in order to verify whether the designed nasal sensor is able to pick up these changes in breathing with respect to RIP. Figure 5a, b show these different respirations and can be seen clearly that both nasal sensors (right and left nostril) picked up the changes in respiration identically. In order to see the level of agreement between the nasal sensors and the RIP device in picking up the alteration in RP, an event (respiration alterations >5 s) has been defined and 2×2 table was constructed. The κ value computed from Table 1 is 0.815. It falls in the range of 0.81–1, which is in good agreement with the data reported elsewhere [25]. This agreement shows that the designed nasal sensor can be used to detect the airway obstruction. Berry et al. [33] have compared the pyroelectric property of the PVDF film with "gold standard" pneumotograph for sleep apnea application. They observed that a PVDF film attached on the lip can detect the apnea as accurately as the gold standard pneumotograph. Therefore, the newly developed nasal sensor reported in this paper can also be used for detecting sleep apnea but the main limitation lies in the positioning of the nasal sensor. As it would be somewhat difficult to sleep while wearing a spectacle frame, an alternative design has to be made for placing the nasal sensor suitably while sleeping. Since this is a preliminary approach to see whether the piezoelectric property of PVDF film can be used to monitor the air flow, the present design has not included the considerations for sleep apnea. However, scope exists for monitoring the patients with sleep apnea with minor design modification.

Few studies have been reported on the use of PVDF film to monitor respiratory parameters. Dodds et al. [14], Huang et al. [18] and Kulkarni et al. [19] have earlier measured the respiration rate by placing PVDF film inside the facemask utilizing PVDF's pyroelectric property. This attachment can affect the respiratory activity and sometimes critically ill patients may not be able to tolerate a facemask [15]. Furthermore, Folke et al. [3] also states that the facemask can even affect respiratory activity such as decrease in respiratory rate, increase in tidal volume and increase in CO_2 production. Siivola [20] and Niizaki et al. [21] have reported about placing PVDF film under bed mattress to record respiration. However in these studies, the authors have mentioned that it becomes difficult to monitor respiration when the subjects have too much of body movements and also sleep posture degrades the respiratory signal quality. Other than these methods, PVDF films have also been used in wearable chest belt sensor for measuring respiratory cycle by Karki and Lekkala [22] and Choi and Jiang [23]. However, by using the chest belts, the respiratory signals may be affected with the cardiac rhythms and a very strong algorithm is required to differentiate the respiratory signal from cardiac signal. In our present study, we have utilized the piezoelectric property of a PVDF film in cantilever configuration which is very simple in design and the sensors were placed freely without any mask or belts. The signal obtained from the designed PVDF nasal sensor is purely due to respiration and do not need any complex algorithm to analyze the signal. There are many other RR monitoring devices using airflow sensing method and they monitor the RR by sensing the expiratory air temperature, humidity, acoustics, CO_2 measured in front of nasal/oral region [3]. Most of these methods use some kind of collector for air, like facemasks or mouthpiece attached to the airways of the subject. The presently designed PVDF nasal sensor is not attached to the airways of the subject and is non-invasive, more precisely contactless (i.e., it is not kept inside the nasal/oral region) unlike nasal prongs, optical sensors, acoustic sensors, nasal thermocouples/thermistors which are placed very close to the airway opening (i.e., just inside the nostrils) [27]. Although the other above mentioned RR measurement methods are non-invasive in nature but they come in contact with the subject which might affect the respiratory activity as explained by Folke et al. [3]. Therefore by using PVDF nasal sensor, the RR can be measured in a natural way without any extra effort from the subjects.

Rhino-manometer is used for detection of the nasal obstruction in the nasal cavity [34], but it is relatively much expensive. In our present nasal sensor design, air flow from each nostril impinges on the nasal sensor giving rise to voltage accordingly. If any of the nostrils is blocked completely or partially, then the air flow from that nostril is less. Correspondingly the amplitude of voltage signal of that nasal sensor is less. This gives quantitative information about the blockage of that particular nostril. During data collection, one of the subject had partial blockage in his left nostril. So there was very less air flow from that nostril. The left nasal sensor has picked up this partial nasal cavity blockage very clearly as seen in Fig. 8, whereas air flow from right nostril and RIP sensor was normal. Therefore using the developed PVDF nasal sensor, the complete/partial blockage in either of the nostril can be found out by calculating the depth or amplitude of the respiration signal from each of the nostril. This can be a potential area of application for the presently developed nasal sensor. The aspect of nasal airflow obstruction is briefly described in this paragraph using the PVDF nasal sensor, although a detailed clinical test is yet to be performed to explore its applicability in this regard. Apart from assessing nasal obstruction, the developed PVDF nasal sensor can also be used to detect airways obstruction, which is far more important clinical information needed. Generally certain airways obstructions like asthma, sleep apnea and COPD cause different types of breathing pattern like tachypnea (RR > 30), bradypnea (RR < 12) and apneas (cessation of breath for an indeterminate period). Also at rest, either increase or decrease in RR decides the need of mechanical ventilation for the patient [4]. As can be seen in Fig. 5a, b, a PVDF nasal sensor was able to pick up different breathing patterns (normal, fast and slow) enabling its applicability to assess the airway obstruction. However, a detailed clinical test is yet to be performed to explore this possibility. The developed PVDF nasal sensor monitors the respiratory parameters like RR, respiratory air flow pattern and respiration magnitude/depth. These parameters could be applied in clinical situations for detecting disorders like sleep apnea, deviated nasal septum and monitoring clinically ill patients. However, detailed studies should be made using PVDF nasal sensor in larger populations to expect more clinically significant results. The limitation of our present study is inclusion of only healthy volunteers instead of patients. Healthy volunteers breathe with their nose, while patients can breathe through their mouths too and the designed PVDF nasal sensor may not detect mouth breathing accurately.

In the present study, data was recorded using a data acquisition system, but it can be developed as a complete stand-alone system consisting of microprocessor, display, memory and computer interface. In our further work, proper design will be made to integrate the PVDF nasal sensor assembly so that it can be used during sleep. Also, a detachable PVDF nasal sensor will be designed for infection control purpose.

A simple questionnaire was prepared and given to the subjects who participated in the study, to know the following.

1. Whether they have any difficulty while wearing the spectacle frame mounted with nasal sensor?
2. According to them which sensor (nasal sensor or RIP device) was easier and more comfortable to wear on their body for monitoring RP?

Around 86 % of subjects answered that wearing nasal sensor mounted on spectacle frame was easier and more comfortable during monitoring of RP compared to wrapping the bands around chest and abdomen for RIP device. Based on these observations, it can be mentioned that the developed nasal sensor is more patient friendly compared to RIP device.

5 Conclusions

The developed PVDF film based nasal sensor is simple to design, non-invasive, compact, portable, patient friendly and may be used routinely to monitor RP by the clinicians. Preliminary results show that the developed PVDF nasal sensor performance compares well with a “gold standard” RIP method for monitoring human respiration pattern in a group of healthy awake human subjects. The study of RP in each nostril can provide qualitative information about complete/partial nasal cavity blockage. Thus, the PVDF nasal sensor is a very useful alternative to the other sensors normally used to monitor the RP.

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