RIB CAGE AND ABDOMINAL PIEZOELECTRIC FILM BELTS TO MEASURE VENTILATORY AIRFLOW

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ABSTRACT. Piezoelectric film–based respiratory belts are described and tested. To the extent that a two-degree of freedom model consisting of rib cage and abdominal motion is able to assist in quantifying ventilation, the piezoelectric belts can monitor flow in a manner analogous to the monitoring of volume with two magnetometer pairs or two respiratory inductive plethysmograph belts. The piezoelectric belts are shown to measure flow linearly when compared with a screen pneumotachometer to a flow of at least 2.6 L/s. There is no phase shift between the peak flow of belts and the pneumotachometer up to a frequency of at least 9.2 Hz. During normal ventilation, 68 and 95% of the peak flows measured with the belts fall within ±10 and 20%, respectively, of the flows measured with a screen pneumotachometer.

KEY WORDS: Measurement techniques: plethysmography. Monitoring: ventilation.

In medical diagnosis and in physiologic evaluation it is desirable to precisely quantify, over time, ventilatory flow. The usual methods involve a mouthpiece or a mask. These devices are not always tolerated by subjects, and, even if tolerated, they interfere with ventilation, the very variable being measured [1]. In addition, mouthpieces and masks limit freedom of movement and are inconvenient and uncomfortable, particularly when used over extended times.

Konno and Mead [2] suggested a way to overcome these problems by quantifying ventilatory volume changes by measuring the movement of the rib cage and abdomen as a reflection of changes in lung volume. Two pairs of magnetometers have been placed on the surface of the rib cage and abdomen to quantify anterior-posterior distance change. The magnetometer pairs produce a voltage proportional to their separation [3].

In addition, a technique has been described to measure the movement of the rib cage and abdomen using insulated wire, which closely encircles the torso and whose inductance changes proportionally to the cross-sectional area enclosed [4]. This variable inductance is connected to a fixed capacitor forming an oscillator circuit whose output frequency is determined by the changing inductance. The variable frequency is converted to a voltage proportional to the cross-sectional area of the torso (respiratory inductive plethysmograph, RIP).

In both of these techniques the two voltages (one each from the rib cage and the abdomen) must be calibrated, that is, they are properly weighted and summed to give a voltage proportional to instantaneous lung volume. The mathematical formulation of the calibration is in

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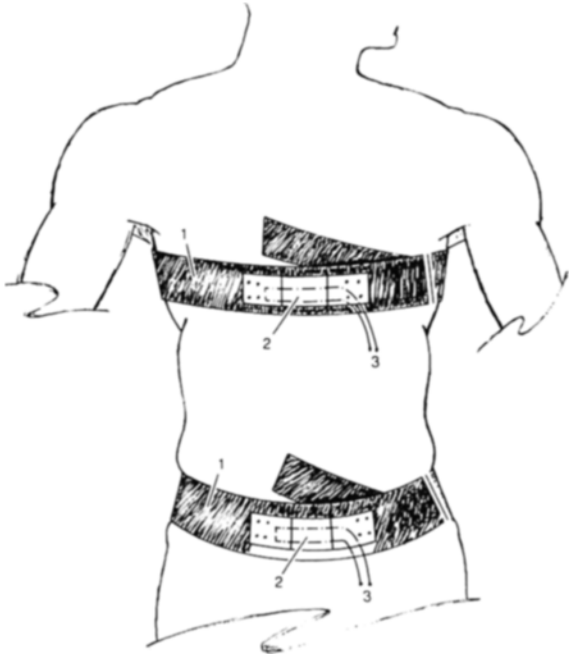


Fig 1. Two stretchable belts at the level of the rib cage and abdomen to quantify respiratory airflow based on a two-degree of freedom system. 1 = stretch Velcro (loops); 2 = piezoelectric film sensor; 3 = electrical connection.

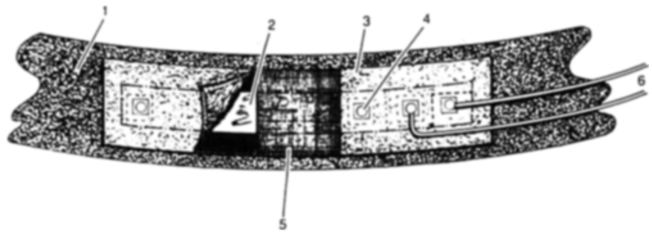


Fig 2. Detailed construction of piezoelectric sensor assembly: 1 = stretch Velcro belt; 2 = piezoelectric film; 3 = Velcro "hooks" tab; 4 = mechanical rivet attachment; 5 = Lycra protective cover; 6 = electrical connection.

both instances based on the description of Konno and Mead [2], who have shown that ventilatory volume can be described by the equation,

$$V = aVRC + bVAB, \tag{1}$$

where V = ventilatory volume, VRC = rib cage excursion, VAB = abdominal excursion, a = rib cage weighting factor, and b = abdominal weighting factor. Magnetometers measure the excursions as anterior-posterior diameter changes and the RIP measures excursions as cross-sectional area changes.

If equation 1 is mathematically differentiated, it becomes

$$\dot{V} = a\dot{V}RC = b\dot{V}AB, \tag{2}$$

where $\dot{V}$ = rate of change of volume or respiratory flow, $\dot{V}RC$ = rate of change of rib cage excursion, and $\dot{V}AB$ = rate of change of abdominal excursion.

The mathematical formulations used for a two-degree of freedom system relating volume to dimensional excursion apply equally to rate of change of volume (flow) relative to the rate of change of excursion.

We describe an alternative measurement technique that uses two elastic belts with series piezoelectric film strips to measure the rate of change of the circumference of the rib cage and abdomen.

Within the limitations of the validity of a two-degree of freedom assumption [5,6], a calibrated summed signal derived from rib cage and abdominal piezoelectric belts retains calibration within a reasonable range over time, with removal and replacement, and with limited upper body motion. The calibration can be performed in an erect or supine subject. Ventilatory flows derived from the belts after calibration agreed in the 2 subjects tested with those measured simultaneously with a screen pneumotachometer.

METHODS

The basis of the device involves elastic belts onto which is attached a polymeric piezolectric film based on polyvinylidene fluoride (Kynar, Pennwalt Corp, Valley Forge, PA). The film is flexible, lightweight, and strong. When stressed by the stretch of the belt, the film produces a voltage proportional to the rate of change of torso circumference. The belts are worn around the rib cage and the abdomen, just as described with the inductive belts of the RIP and as shown in Figure 1. Each of the two belts measures one degree of the two-degree of freedom system described by Konno and Mead [2].

Belt Construction

The belts are made in the author's laboratory. The sensor part of the belt is shown in Figure 2. The Kynar film (22 mm × 80 mm × 52 μm) is sandwiched on both ends between short tabs of Velcro hooks. Lycra is sewn between the Velcro tabs as a protective cover for the Kynar. The mechanical coupling between the Kynar and the Velcro is made by adhesive cement and plastic rivets through the Kynar and Velcro material. Electrical connection to the silver ink surface of the Kynar is made by pass-through metal rivets.

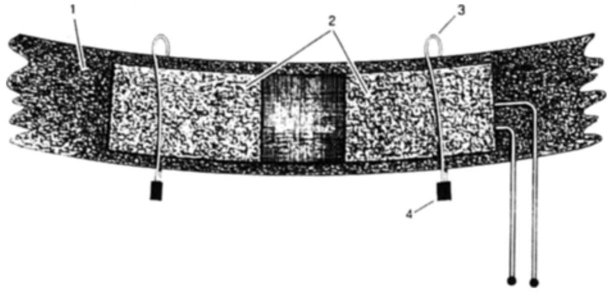


Fig 3. Attachment of sensor assembly to stretch belt: 1 = stretch belt; 2 = Velcro "hooks" tab; 3 = spring clips; 4 = rubber reinforcement sleeve.

The sensor assembly attaches to the elastic Velcro belt "hooks" on the sensor to "loops" of the belt, as shown in Figure 3.

During periods of extended movement, as might occur during sleep, the Velcro binding may slide under the sheer stress, and the static tension on the sensor may be diminished. The binding is reinforced by spring clips to prevent sliding.

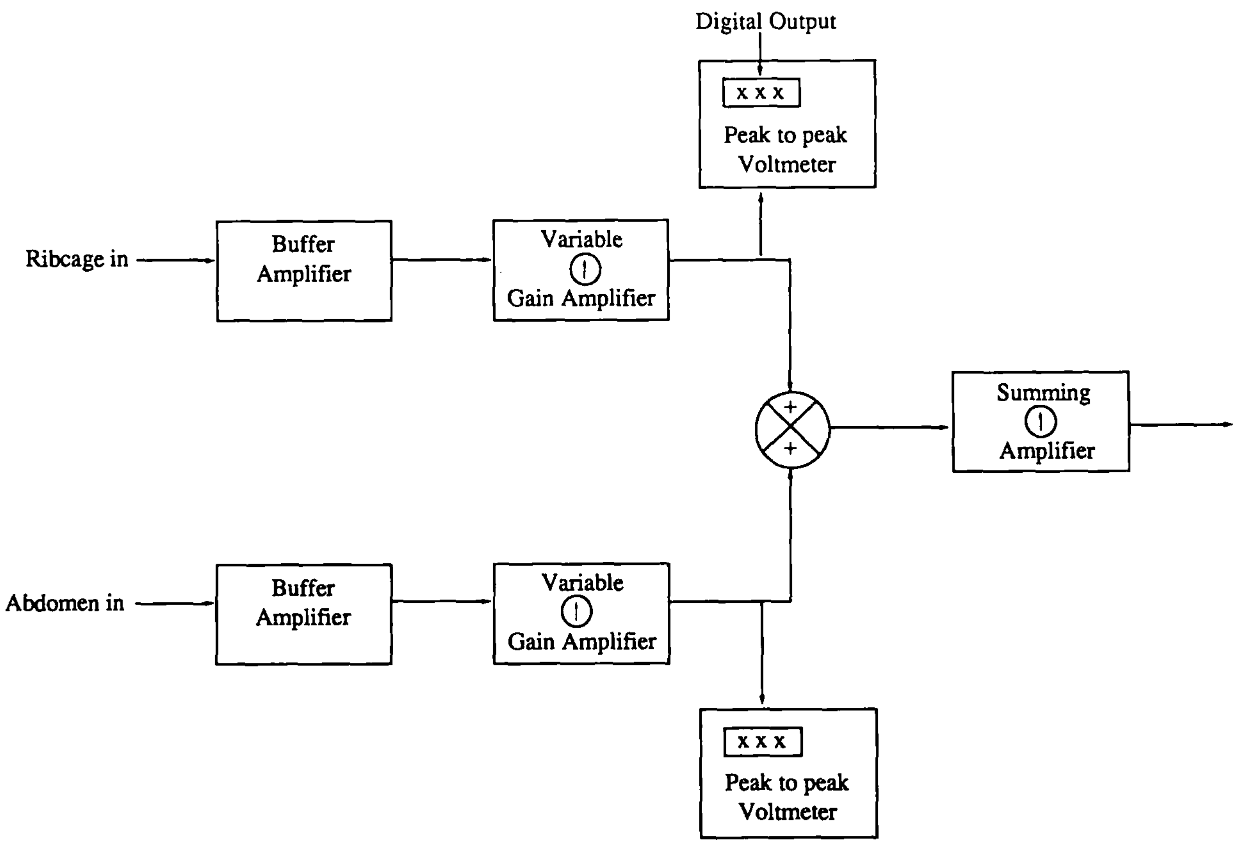
Electronic Processing

Figure 4 is a block diagram of the electronic processing of the voltage signals from the two piezoelectric belts. The voltage generated by each belt is buffered by an amplifier with input impedance of 1 MΩ and then amplified by a variable gain amplifier. The outputs of the two amplifiers are summed in a summing amplifier. The averaged (2 seconds) peak-to-peak amplitude is displayed. This averaged peak-to-peak amplitude will be used for isovolume calibration, as described in the next section. This electronic processor was manufactured by Buxco Electronics, Inc, Sharon, CT.

There are several potential sources of noise on the flow signal:

1. Interference associated with 60-Hz (or other frequency) electrical noise has not been found as yet.
2. Substantial noise is associated with motion of the person being monitored. This noise signal can be greater than the desired ventilatory signal. Fortunately the noise disappears when motion disappears and is much different in character from the ventila-

Fig 4. Electronic processor for amplification and calibration of the sensor output voltages.



tory signal (high-amplitude, high-frequency bursts), so that it is easy to distinguish from the ventilatory signal, although the monitoring of ventilation is not possible during periods of movement. If automatic signal processing is desired, these artifacts will need to be filtered, just as with the RIP [7].

3. Noise is associated with cardiac activity. During ventricular contraction and filling, the ventricle pulls and pushes on the lungs, causing rib cage and abdominal contraction and expansion. This is sensed by the piezoelectric belts as a "noise" signal on the respiratory signal. The signal–to–"cardiac" noise ratio can vary from 50:1 to 2:1, and if desired can be reduced by passing the signal through a 3-Hz low-pass filter. This, of course, will produce a phase shift with the "true" signal.

Calibration

Calibration, as performed in 10 subjects, is based on the isovolume method described by Konno and Mead [2]. There are two steps for calibration. First, the relative gains of each of the belt amplifiers are set so that the summed signal is proportional to the subject's total airflow. Next, the gain of the summing amplifier is set so that the summed signal from the belts equals the airflow signal as measured by a pneumotachometer or an equivalent airflow measuring device.

A specific procedure was followed for the studies. The protocol for these studies was approved by the Hospital Institutional Review Board. The subject was asked to perform an isovolume maneuver, which required a cooperative subject. Ten consecutive naive subjects were all able to perform the isovolume maneuver when given the following instructions:

> Create a gentle inward and outward motion with your abdomen while your mouth and nose are blocked. This will shift air back and forth from the top of your lung to the base. Repeat the maneuver six to ten times.

This maneuver was demonstrated and then attempted by each subject. An average of eight attempts with continual feedback was necessary before the manuever was mastered in naive subjects. The subject concentrated on creating abdominal motion rather than ventilation as the primary effort.

When the subject was able to successfully perform the isovolume maneuver, the envelope of peak-to-peak voltage (or the 2-second average) of the two signals, rib cage and abdomen, was adjusted to be equal. The gain of the summing amplifier was adjusted to be equal to the previously calibrated pneumotachometer signal while the subject breathed normally through the pneumotachometer.

Testing

INDIVIDUAL SENSOR (LINEARITY AND FREQUENCY RESPONSE). A piezoelectric belt with its sensor was placed around an inflated 3-L anesthesia bag. The bag was inflated and deflated at varying rates by pumping a resuscitation bag through a screen pneumotachometer. The output of the sensor around the bag and the output of the pneumotachometer were recorded on a Model 78 EEG Polygraph Data Recording System using 7P122 DC amplifiers (Grass Instrument Co, Quincy, MA). The resistance of the screen pneumotachometer and anesthesia bag tubing connector was 0.4-cm $H_2O \cdot L \cdot s$. The compliance of the anesthesia bag was 0.1 L/cm H_2O.

INDIVIDUAL SENSOR (CIRCUMFERENTIAL HOMOGENEITY). Four sensors were attached at 90° separation to a single belt around the abdomen or the rib cage and the four outputs were monitored simultaneously in an erect subject. Identical signals imply that stretch on the belt at the four sites is reflected identically as stress on the sensor.

COMPARISON OF SUMMED FLOW WITH SCREEN PNEUMOTACHOMETER FLOW. After calibration, ventilatory flow as measured by the summed piezoelectric belts was compared with that from a screen pneumotachometer. The ratio of the rib cage to abdominal contribution to ventilation was varied to exaggerate potential calibration errors.

REPRODUCIBILITY OF CALIBRATION DURING ISOVOLUME MANEUVERS. To test the constancy of calibration under conditions simulating potential clinical uses of the piezoelectric belts, several tests of reproducibility were performed in 2 subjects.

First, to document the reproducibility of the isovolume maneuver, 2 subjects repeated the isovolume maneuver five times after an initial calibration.

To test the sensitivity of calibration to position, placement, and tension of the belts, the isovolume calibration maneuvers were repeated following momentary removal and replacement of the belts.

To simulate the use of the piezoelectric belts in polysomnography, the subjects performed the isovolume calibration while supine. The calibration was checked with the subject still supine.

Finally to test the reproducibility of calibration under conditions in which the subject moves, the calibration

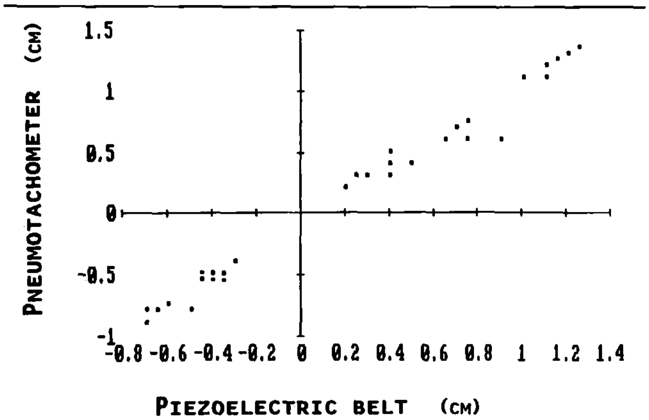


Fig 5. Data obtained from an anesthesia bag model. The peak rate of change of circumference is plotted against simultaneously measured pneumotachometer flow in and out of the bag. The correlation coefficient is 0.993. Axes are labeled in centimeter deflection, with 1 cm equal to 2 L/s on the pneumotachometer axis.

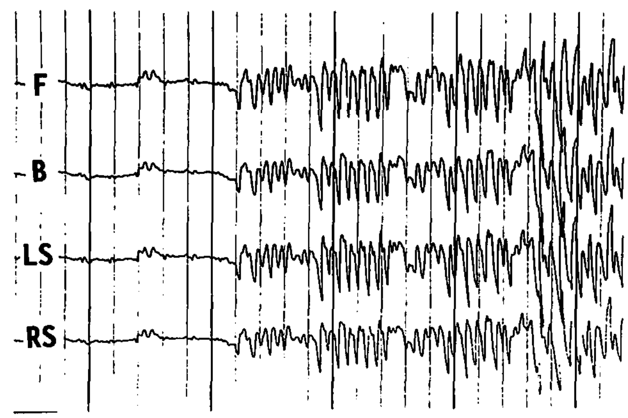


Fig 6. Simultaneous measurement of rib cage motion from a sensor assembly on the front (F), *back* (B), *and left* (LS) *and right* (RS) *sides during a deliberately varying respiratory pattern recorded at 10 mm/s.*

was checked after the subjects clapped their hands above their heads ten times.

NORMAL VENTILATION. The performance of the RIP was examined by comparing tidal volumes as measured with the RIP with the tidal volumes measured simultaneously with a spirometer [4]. In an analogous fashion, we compared the peak-to-peak respiratory flows as measured with the piezoelectric belts with the flows measured simultaneously with a pneumotachometer. The measurements were performed on 4 subjects during normal breathing. At least 40 consecutive breaths were analyzed for each subject.

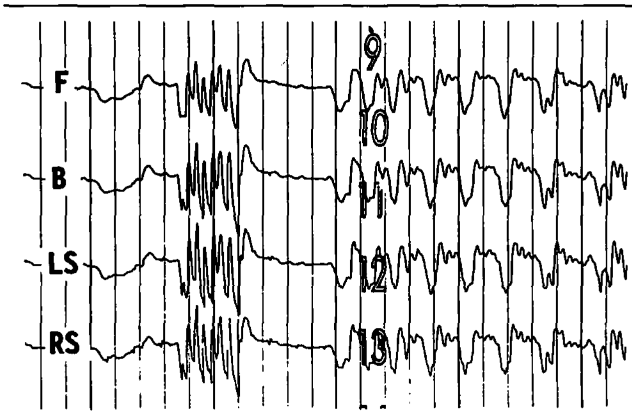


Fig 7. Simultaneous measurement of abdominal motion from a sensor assembly on the front (F), *back* (B), *and left* (LS) *and right* (RS) *sides during a deliberately varying respiratory pattern recorded at 10 mm/s.*

RESULTS

Individual Belt (Linearity and Frequency Response)

With the use of an anesthesia bag as a model of the thoracic cage, peak flow recorded from the piezoelectric belt was plotted versus the peak flow from the pneumotachograph (Fig 5). The range of flow was −1.8 to 2.6 L/s. The correlation coefficient was 0.993. The time delay between peak flow of the piezoelectric belt and pneumotachometer was constant at 0.033 seconds at frequencies from 1.5 to 9.2 Hz.

Individual Belt (Circumferential Homogeneity)

Figures 6 and 7 demonstrate the respiratory flow signals generated from four sensors: on the front, on the back, and on the sides of the same belt of a subject. Figure 6 is from a rib cage belt and Figure 7 from an abdominal belt. The signals indicate that stretches generated on belts at the four locations on the circumference of the belt are congruent.

Comparison of Summed Flow and Screen Pneumotachometer Flow

The calibrated, summed signal representing total ventilatory flow for a two-degree of freedom system was compared with the flow measured with a subject breathing through a screen pneumotachometer. The subject

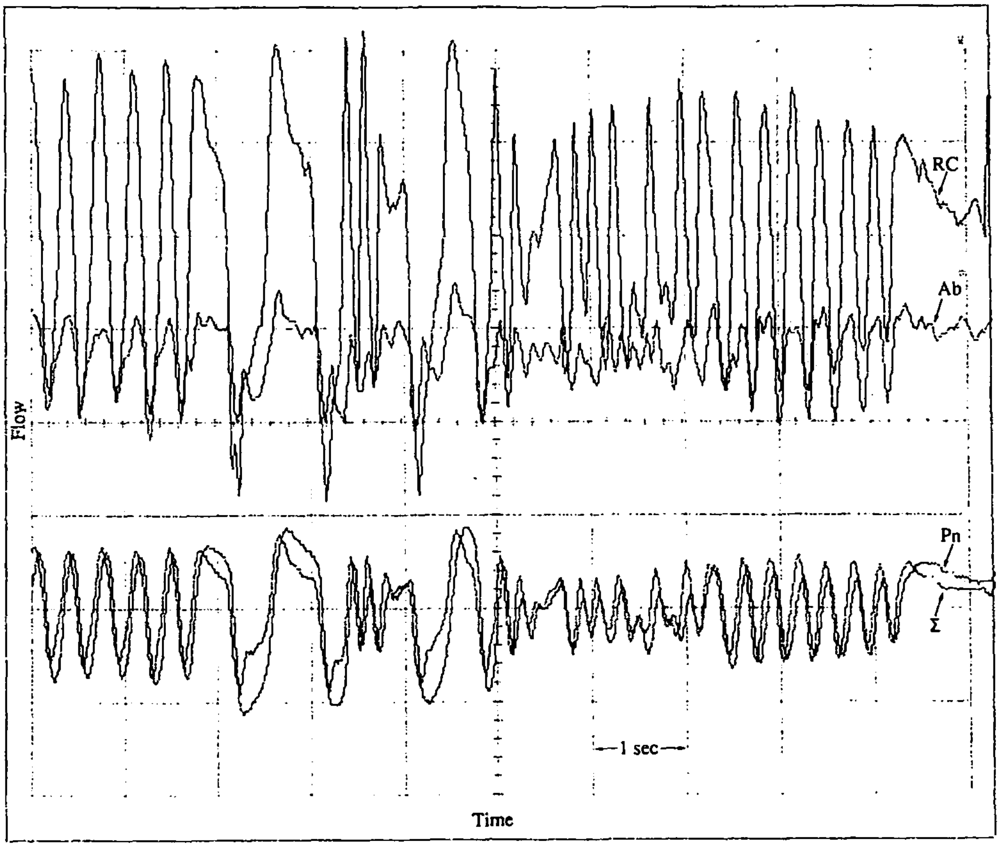


Fig 8. Respiratory flow recorded at 1 s/division. RC = rib cage; Ab = *abdomen;* Σ = *weighted sum of rib cage and abdomen;* Pn = *pneumotachograph. Expiration is up and inspiration down. The y scale is 2 L · s · division.*

had been asked to change his breathing pattern continuously to exaggerate any differences in response of the two measures of flow.

Figure 8 compares ventilatory flows recorded at a speed of 1 s/division. The upper two traces (rib cage and abdomen) suggest that in this respiratory maneuver lung flow was primarily associated with rib cage flow (approximately 3:1). Expiration is up and inspiration down. Instantaneous flow varies from zero to 2 L/s. The pneumotachometer and summed flow signals are similar at all amplitudes. There is a time delay between the two signals. This is caused by gas compression in the lungs and is proportional to specific airway resistance [8]. Minor flow variations during breathing are reflected equally on both traces.

Figure 9 compares the summed and the pneumotachometer flows at a recording speed of 200 ms/division. In the maneuver illustrated, the rib cage and abdominal flow excursions are variable (upper two traces), being anywhere from in phase with each other to 180° out of phase. In all conditions the summed and pneumotachometer signals remain similar, although the difference between the pneumotachometer and summed signal is as much as 30% in this extreme test. Again, the phase shift between the summed and pneumotachometer signals reflects the effect of gas compression in the lung.

Calibration Reproducibility

After an initial calibration, the ratios of calibration factors in 2 subjects were 0.96 ± 0.03 and 0.89 ± 0.03, respectively, when repeated isovolume maneuvers were performed.

With momentary sensor removal and replacement between calibration and validation, the ratios were 1.22 ± 0.16 and 1.40 ± 0.33.

With the subjects in the supine position, calibration factor ratios were 1.0 ± 0.24 and 1.17 ± 0.30 in the 2 subjects.

After the subjects clapped their hands above their head ten times to create upper body motion, the calibration factor ratios were 1.03 ± 0.08 and 1.13 ± 0.09.

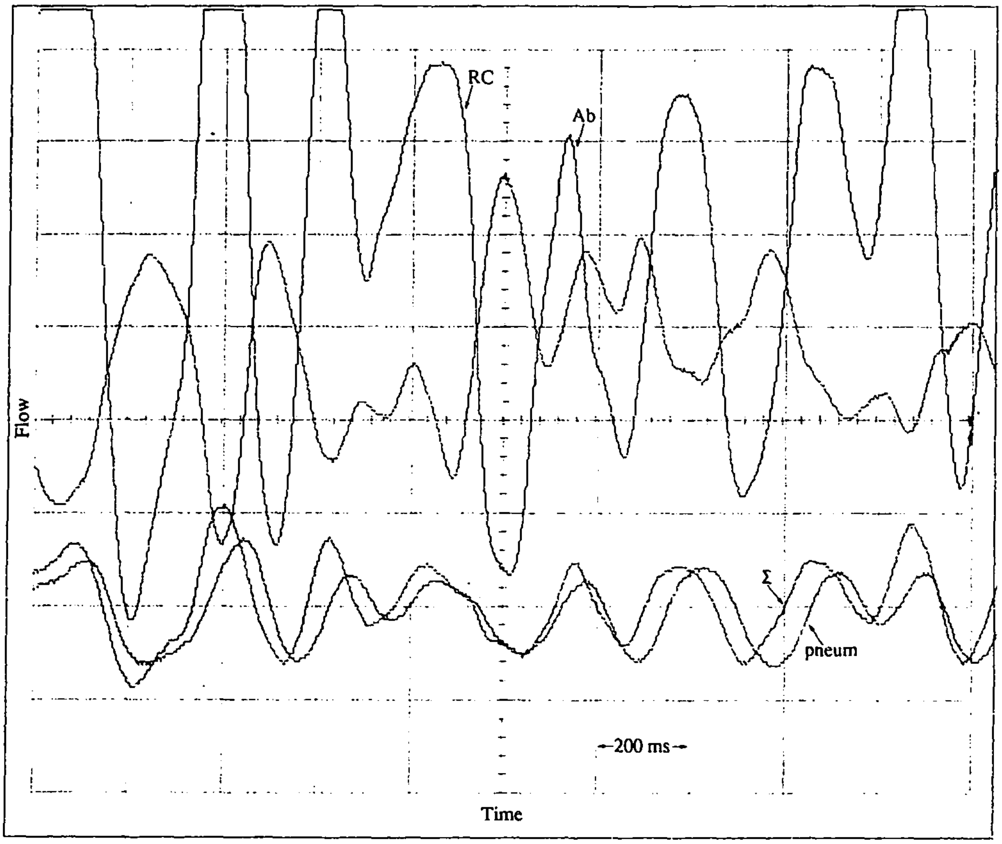


Fig 9. Respiratory flow recorded at 200 ms/division. RC = *rib cage;* Ab = *abdomen,* Σ = *weighted sum of rib cage and abdomen;* pneum = *pneumotachograph. The vertical scale is 2 L · s · division.*

Normal Ventilation

In 4 subjects, five series of 40 to 100 consecutive breaths were analyzed by comparing the difference between peak inspiratory and peak expiratory flow as measured by the calibrated, summed flow signal derived from the piezoelectric belts and as measured with a screen pneumotachometer. The scatter of the percent difference was characterized by standard deviations of 9.0, 6.0, 7.6, 13.6, and 12.4%. The mean standard deviation was 9.7%.

DISCUSSION

Motion of the rib cage and abdomen has been used for many years to qualitatively describe ventilation. Konno and Mead [2] have described a two-degree of freedom system relating instantaneous ventilatory volume to instantaneous rib cage and abdominal expansion. Mathematical differentiation of their describing equation generates an equation that implies that instantaneous ventilatory flow must also be described by instantaneous rib cage and abdominal rate of change of expansion.

Magnetometers and the RIP have been used successfully to provide a nonencumbering quantitative measurement of ventilation. There are some limitations to their use. Some of these limitations are instrument related and some are related to the failure of the two-degree of freedom assumption [5,6]. We have examined the use of a piezoelectric belt system within the limits of a two-degree of freedom assumption.

The major difference in the output signal from piezoelectric film as compared with that from the magnetometers or RIP is that the output is ventilatory flow in the former and ventilatory volume in the latter two. These are, of course, interchangeable by electronically integrating the flow or differentiating the volume. Flow allows easier quantification of fast events, but volume has been a more usual descriptor of ventilation.

Concerning the technical performance of the piezoelectric belt, the response appears to be linear from −1.8 to 2.6 L/s. The response may be linear over a wider range but was not tested. The time delay between peak flow of the belt and pneumotachometer remained fixed at 0.033 seconds at frequencies up to 9.2 Hz. We have not measured the delay above 9.2 Hz. The major part of this delay can be attributed to the time constant of the anesthesia bag system used to quantify this characteristic. The resistance of the inlet (0.4 cm $H_2O \cdot L \cdot s$) times the compliance of the bag (0.1 L/cm H_2O) is 0.04 seconds. There is also a time delay of about 0.01 seconds in the response of the piezoelectric sensor. This small delay varies as a function of the surface area of the piezoelectric film and the input impedance of the buffer amplifier.

The piezoelectric film was able to measure circumference change when it was placed anteriorly, posteriorly, or laterally.

The similarity of the signals derived from the belt to those derived from the screen pneumotachometer suggests that the belts, even with a two-degree of freedom assumption, are able to detect fine details of ventilation. The clinical implications of this capability have not been explored, although it might be useful for measurement of specific airway resistance [8].

Calibration was examined by measuring the ratio of the calibration factors during a repeated isovolume maneuver after an initial calibration. A ratio of one corresponds to an isovolume line angle [2] of 45°. In measurements validating the RIP, Cohn et al [4] found that 50% of their isovolume line angles were within ±3° of 45°. This corresponds to a calibration factor ratio of 0.90 to 1.11. The ratios in our 2 patients were very close to being within this range with repeated measurements, whether erect (0.89, 0.96), supine (1.0, 1.17), or after upper body motion (1.03, 1.13). After removing and replacing the belts, the calibration ratios were 1.22 and 1.4, corresponding to isovolume line angles of 50.6° and 54.5°, respectively.

Comparing ventilatory flow as measured by the summed belt and a pneumotachometer, the standard deviation of the percent difference between the two was approximately 10%. Cohn et al [4] have found that 73 and 91% of the slow vital capacities measured by the RIP fall within ±10 and 20%, respectively, of the volumes measured by spirometry. The 10% standard deviation that we found implies that 68% and 95% of the peak flows measured by piezoelectric film fall within ±10 and 20%, respectively.

In summary, a piezoelectric film–based respiratory belt has been described and tested. It produces a signal that is proportional to flow. This signal is linear. Belts may be used in pairs in a two-degree of freedom arrangement to monitor total ventilatory flow.

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