Phonospirometry for noninvasive measurement of ventilation: methodology and preliminary results

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Que, Cheng-Li, Christof Kolmaga, Louis-Gilles Durand, Suzanne M. Kelly, and Peter T. Macklem. Phonospirometry for noninvasive measurement of ventilation: methodology and preliminary results. J Appl Physiol 93: 1515–1526, 2002. First published June 14, 2002; 10.1152/japplphysiol.00028.2002.—We measured tracheal flow from tracheal sounds to estimate tidal volume, minute ventilation (Vt), respiratory frequency, mean inspiratory flow (Vt/TI), and duty cycle (Vt/Ttot). In 11 normal subjects, 3 patients with unstable airway obstruction, and 3 stable asthmatic patients, we measured tracheal sounds and flow twice: first to derive flow-sound relationships and second to obtain flow-volume relationships from the sound signal. The flow-volume relationship was compared with pneumotach-derived volume. When subjects were seated, facing forward and with neck extension and while supine. We then measured ventilation without mouthpiece or nose clip from tracheal sounds during quiet breathing for up to 30 min. Normal results ± SD revealed tidal volume = 0.37 ± 0.065 liter, respiratory frequency = 19.3 ± 3.5 breaths/min, Vt = 6.9 ± 1.2 l/min, Vt/TI = 0.31 ± 0.06 Vt/s, and Vt/Ttot = 0.37 ± 0.04. Unstable airway obstruction had large Vt due to increased Vt/TI. With the exception of Vt/Ttot, variations in ventilatory parameters were closer to log normal than normal distributions and tended to be greater in patients. We conclude that phonospirometry measures ventilation reasonably accurately without mouthpiece, nose clip, or rigid postural constraints.

Minute ventilation; pattern of breathing; breath sounds; tidal volume; breathing frequency

Precise monitoring of ventilation is usually achieved by having a subject breathe through a mouthpiece or face mask attached to a pneumotachygraph or spirometer. Although these devices permit the accurate measurement of ventilation and its parameters, they also alter the pattern of breathing and minute ventilation (Vt) (2, 12, 13, 37). They are not useful for monitoring ventilation in any circumstance in which keeping a mouthpiece and nose clip in place is too difficult or impossible.

Magnetometry and inductance plethysmography record the movements of the rib cage and abdomen during respiration and, by the use of a suitable calibration procedure, convert the summed thoracic and abdominal motions into a volume signal (30, 34, 35). Although these devices eliminate the need for a mouthpiece and nose clip and, as a result, permit measurement of the volumes, flows, and timing of normal breathing, they are unfortunately sensitive to changes in posture, particularly xyphi-pubic distance (24, 34, 35, 38, 39). Even when the latter is controlled, the rib cage and abdomen do not always act as compartments with a single degree of freedom (7, 14, 21, 36), and this introduces a further source of error. Motion analysis by optoelectronic plethysmography works extremely well (5) but is quite expensive and constrains the subject because markers on the trunk must be visualized by fixed videocameras. There is a need for other devices that can measure ventilation cheaply and accurately without a mouthpiece and nose clip.

Phonospirometry, the estimation of ventilation parameters from measurements of tracheal breath sounds, provides a simple alternative to motion analysis systems and may prove to be more versatile. Since the invention of the stethoscope by René Laennec in 1819, auscultation has provided the clinician with a quick, but crude, assessment of pulmonary ventilation. Objective measurements of breath sounds were first made more than 25 yr ago (17) and have been used as a qualitative assessment of ven-
tilation during sleep (8); but it is only with advances in computer technology and the wide application of digital signal analysis that recording and analysis of respiratory sounds has accelerated.

Normal breath sounds are primarily generated by turbulence within large- and medium-sized airways. Airflow velocity and airway dimensions influence the generation of turbulent flow, whereas the intensity of the detected sound is influenced by the sound transmission characteristics of the tissue between the regions of sound generation and the point where measurements are made. As a result, for a given subject and microphone position, sound amplitude is proportional to the flow rate (10, 15, 17, 32, 33), suggesting the possibility of deriving flow estimates from sound amplitude (33). In this paper, we present a new method to measure ventilation by converting tracheal breath sound intensity to ventilatory flow. By integration, tidal volume (VT), respiratory frequency (f), Vm, and other ventilatory parameters are obtained.

Many investigators have measured airflow by acoustical techniques (1, 6, 10, 18, 19, 27, 28). However, their interest has lain primarily in the derivation of parameters related to the frequency spectrum of the sound signal, or the mechanism of sound production, rather than in the use of the sound signal as a surrogate for flow. With the exception of Soufflet et al. (33), the studies quantifying the relationship between sound intensity and flow (10, 32) have focused exclusively on flows of >0.5 l/s.

Our purpose was to estimate flow from breath sound intensity during quiet breathing and, therefore, almost exclusively on flows of <0.5 l/s. This presented particular problems of an adverse signal-to-noise ratio (SNR) and the threshold of flow below which flow-derived sound cannot be detected. These problems have not previously been satisfactorily addressed. We have solved these problems by measuring the threshold of detection of flow-derived sounds (∼0.3 l/s) and interpolating from the threshold to zero-flow points. Thus we were able to measure Vr, f, inspiratory (Ti) and expiratory (Tc) durations, respiratory cycle time (Ttot), duty cycle (Tc/Ttot), and mean inspiratory flow (Vr/Ti) during quiet breathing from the envelope of the sound signal with only a microphone placed on the skin over the thyroid cartilage in the neck, without the use of a mouthpiece or nose clip except for a calibration procedure. In this paper we describe this technology and present preliminary results.

**METHODS**

**Subjects.** A total of 11 nonsmoking normal volunteers with no history of pulmonary disease or recent respiratory tract infection, 3 asymptomatic mild asthmatic patients, and 3 patients with unstable airway obstruction attending the day hospital of the Montreal Chest Institute were recruited for the study. Of these, seven normal subjects and one asymptomatic asthmatic subject participated in the early studies in which we validated phonospirometry as a method to measure ventilation, estimated the errors by a Bland-Altman analysis, and tabulated the ventilatory pattern shown in Table 1. SNR and the power spectrum of normal breath sounds were measured in one of these subjects and in four additional normal volunteers. Effects of posture were studied in 5 of the 11 normal subjects. Subsequently five additional patients, two with mild asthma and three with unstable airways obstruction, attending the day hospital of the Montreal Chest Institute, were studied to measure their ventilatory pattern and its variations. These patients were studied primarily to get a feeling for how applicable the technique is to disease rather than to make conclusions as to the effect of disease on breathing pattern.

**Calibration procedure.** Airflow at the mouth and tracheal breath sounds were measured simultaneously during two separate 30-s intervals. The setup is illustrated in Fig. 1. Subjects were seated, wore a nose clip, and breathed quietly on a mouthpiece. An electret microphone (model 1306, Armaco, Vancouver, Canada), coupled to a custom-designed preamplifier with a 30-fold amplification, was placed over the trachea at the level of the thyroid cartilage, a region that had previously been determined to provide the best sound signal. The microphone was inserted in a conical aluminium housing 5.5 mm in diameter around the microphone and 13 mm in diameter at the opening that was placed on the skin. The distance between the microphone and the opening was 3.9 mm. Frequency response of the electret microphone was flat (±3 dB) between 20 Hz and 8 kHz. Flow signal was obtained by using a pneumotachygraph (model no. 1A, Fleisch, Lausanne, Switzerland) and a piezoelectric pressure transducer. Both flow and sound were amplified and filtered before analog-to-digital conversion. For flow, the cutoff was set at 50 Hz; for the sound signal, cutoff was at 1,000 Hz. Both signals were sampled at 3,000 Hz by using a commercially available software package (ORIGIN, MICROCAL, Northampton, MA) and a 12-bit analog-to-digital converter (model DT-2891, Data Translation, Marlboro, MA).

For one 30-s period, the relationship between flow and sound was determined (see below). This relationship was

Table 1. **Effect of head position on error of Vr estimation**

<table>
<thead>
<tr>
<th>Subject</th>
<th>Forward</th>
<th>Left</th>
<th>Right</th>
<th>Up</th>
<th>Down</th>
</tr>
</thead>
<tbody>
<tr>
<td>SY</td>
<td>0.08 ± 0.08</td>
<td>0.16 ± 0.11</td>
<td>0.23 ± 0.10</td>
<td>0.14 ± 0.07</td>
<td>0.10 ± 0.08</td>
</tr>
<tr>
<td>JH</td>
<td>0.12 ± 0.06</td>
<td>0.11 ± 0.05</td>
<td>0.11 ± 0.08</td>
<td>0.26 ± 0.19</td>
<td>0.12 ± 0.05</td>
</tr>
<tr>
<td>CL</td>
<td>0.15 ± 0.10</td>
<td>0.15 ± 0.05</td>
<td>0.14 ± 0.07</td>
<td>0.21 ± 0.11</td>
<td>0.16 ± 0.15</td>
</tr>
<tr>
<td>QL</td>
<td>0.09 ± 0.07</td>
<td>0.09 ± 0.09</td>
<td>0.03 ± 0.01</td>
<td>0.28 ± 0.23</td>
<td>0.15 ± 0.09</td>
</tr>
<tr>
<td>EB</td>
<td>0.08 ± 0.06</td>
<td>0.11 ± 0.07</td>
<td>0.16 ± 0.05</td>
<td>0.07 ± 0.06</td>
<td>0.07 ± 0.05</td>
</tr>
<tr>
<td>SK</td>
<td>0.07 ± 0.05</td>
<td>0.07 ± 0.01</td>
<td>0.12 ± 0.03</td>
<td>0.16 ± 0.12</td>
<td>0.10 ± 0.08</td>
</tr>
</tbody>
</table>

Values are means ± SD. Subjects are identified by their initials. Vr, tidal volume; Ve, VT estimated from breath sound intensity; Vm, VT measured by integration of flow.
then used to derive flow and volume from the amplitude of the sound signal of the second measurement period. We assessed the accuracy of this acoustic method by comparing the volume obtained from integration of the measured flow (Vm) to the flow estimated from the sound signal (Ve). This calibration procedure was done individually for each subject in the seated position. When it was successfully completed, breath sounds alone were used to estimate flow and the ventilatory parameters were derived from this.

Effect of neck rotation and body posture on volume measurements. Because head movements and changes of posture are to be expected during long-term measurements of ventilation, we examined the correspondence between Vm and Ve signals during systematic changes of head and body position. First, in five normal subjects and one asymptomatic asthmatic subject (SK), flow and sound signals were obtained during up-and-down and right-to-left movements of the head. Subjects were asked to breathe quietly for 10 s while facing forward, for 20 s while facing left and right, respectively, and then for a final 10 s facing forward for a second time. Second, maximum up-and-down movements of the head were studied in a similar fashion, i.e., a subject breathed quietly for 10 s with the head positioned centrally, for 20 s each with the neck maximally extended then flexed, then finally for 10 s facing forward. At a f of 15 breaths/min, 20 s would be sufficient to record about five breaths; in fact we found that f was closer to 20 breaths/min so that with rotation, extension, and flexion we measured between six and seven breaths.

To determine whether the calibration made seated could be used in different postures, recordings of sound and flow were obtained during two 30-s periods in each of three postures: sitting, standing, and supine. We determined how accurately Vt calculated from the sound envelope using the calibration made while seated estimated Vt measured in the standing and supine postures by integrating flow. In addition, the reproducibility of the calibration was verified by repeating the measurements in the seated position at the end of the experiment.

Data analysis. Figure 2 is an example of the data obtained from one subject during a 30-s calibration period. Only a part of the record is illustrated. Shown from above downward are flow, corresponding sound signal, filtered sound signal, and sound envelope. During a single breath, two major sound bursts are observed (Fig. 2B), one during inspiration, the other during expiration. Sounds in phase with the heartbeat are seen as sharp transients. The digitized sound signal was band-pass filtered between 200 and 1,000 Hz to remove heart and muscle sounds, which are typically <200 Hz, and the high-frequency noise, which is >1,000 Hz. Specifically, the digitized data was transformed to the frequency domain by using a discrete Fourier transform (DFT). Then, all negative frequencies <1,000 Hz, and frequencies between −200 and 200 Hz were set to zero by using a three-point roll-off function. To provide a better attenuation in the rejection bands of the DFT-based band-pass filter, the three first frequency coefficients at the upper and lower boundaries of the pass bands were weighted by the following coefficients: 0.022, 0.23, 0.70 (11). After band-pass filtering (Fig. 2C), there was a marked attenuation of both the background noise and the cardiac artifact, whereas the breath sound signal was well preserved.

To obtain sound intensity, we integrated the sound signal by applying the Hilbert transform to our digital data (23). This was done in conjunction with the frequency-domain filtering mentioned above and required that frequencies between −200 and −1,000 Hz also be set to zero. After an inverse DFT, the resulting envelope signal was filtered in two steps to further reduce noise. First, data was subsampled by averaging the 100 closest neighbors at every 50th data point. Second, a 20-point second-order least-squares smoothing filter was applied. The resulting signal is illustrated in Fig. 2D. The flow signal was also subsampled by a factor of 50 to maintain the same frequency as the sound signal.

In Fig. 3, the relationship between sound amplitude and flow in eight normal subjects is plotted for all breaths of a single measurement period. As found by others (33) below ~0.3 l/s, sound amplitude did not exceed the background noise. A threshold sound value above which the relationship between sound amplitude and flow could be determined was thus defined. For each subject, the threshold was set so as to exceed both the background noise and the transition where tracheal sounds are just apparent. As a result, there was an approximately linear relationship between flow and tracheal breath sound amplitude. Because this relationship was different during inspiration and expiration, these intervals were analyzed separately, and, for each of inspiration and expiration, the relationship between sound amplitude and flow was determined by least squares linear regression.

Because the start and end of each inspiration can easily be determined from the sound amplitude signal (Fig. 2D), flow below the threshold value was estimated by linear interpolation between zero flow points and the first time points when flow could be obtained from the sound signal (Fig. 4). Interpolation fits the data well during inspiration but not during expiration. Therefore, Vt was estimated from inspiratory sounds. As seen in Fig. 4, zero flow points were properly identified.

From the Vt estimates as a function of time and the zero flow points, we calculated f, Vt, Ti, Tt, Vt/Tt and Ti/Ttot. We constructed frequency distributions of all ventilatory parameters before and after log transformation. We measured means ± SD (μ and σ for log-transformed data). Frequency was normalized by expressing it as a fraction of the total number of measurements and for choice of bin width, thus obtaining the probability density functions (PDF). We estimated skewness and kurtosis of the gaussian and log-normal distributions. We used an unpaired t-test to compare ventilatory patterns between normal subjects and patients.

RESULTS

The relationship between sound intensity and flow in seven normal subjects and an asthmatic subject (SK) is illustrated in Fig. 3. The general shape is similar for all subjects, although there were variations in sound intensity at a given flow from subject to subject as well as in the flow value above which tracheal sounds could be detected. In Fig. 4, flow derived from the sound signal (solid lines) is compared with the measured flow signal.
The thick segments represent the calculated flows, and the thin solid lines represent the interpolated values. During inspiration, calculated flow is quite close to the measured value and even reflects some of its details. Expiratory flows are less well estimated from the sound signal and, in this case, underestimate the true value. Similarly, the interpolated values are more accurate during the entire inspiration interval and the beginning of expiration, but the concave shape of the actual flow curve during the latter part of expiration is poorly estimated. This error was greatest when subjects had a long end-expiratory pause.

To determine more precisely how sound intensity changed with flow, in five of the normal subjects shown in Fig. 3 and in SK, we compared the power spectrum of the sound signal obtained during a 15-s breath hold to the spectra obtained during constant flow inspirations between 0.07 and 0.3 l/s. Data from a representative subject is illustrated in Fig. 5A. In this subject, inspiratory sounds were detectable above the background noise at 0.15 l/s and progressively increased in amplitude as flow was increased. Similar results were seen in the other five subjects, but the threshold for inspiratory sound detection differed from individual to individual. SNR was calculated by dividing, at every frequency, the sound amplitude during inspiration by the sound amplitude during the breath hold and then averaging the values for all frequencies between 200 and 1,000 Hz. These values, plotted as a function of the mean flow rate in Fig. 5B, show the intersubject variation in the threshold flow rate above which sound could be detected above background. The threshold

Fig. 2. A segment of quiet breathing from 1 subject. A: flow measured by the pneumotachygraph. B: tracheal sound signal. C: sound signal after band-pass (200–1,000 Hz) filtering. D: sound envelope obtained after filtering and Hilbert transform.
value was generally specific to each subject. Above it was a quasi-linear relationship between SNR and flow in all but one subject.

Measured (solid lines) and calculated (dotted lines) flows for the eight subjects in Fig. 3 are shown in Fig. 6. In subjects CL, QZ, SK, QL, EB, and SW, the two flow signals were very close during both inspiration and expiration, whereas in the remaining subjects, there were significant differences between inspiration and expiration. These differences were greatest during expiration, and, in all subjects, the two signals were very similar during inspiration.

Validation of transfer functions. Because we were more interested in volume than in flow, we used the error in volume estimation as our criterion of accuracy for the technique. The relationship between $V_e$ and $V_m$ is shown in Fig. 7A. The solid line represents the line of identity. All data points fall near the identity line, which indicates a good correspondence between $V_e$ and $V_m$. The breath-by-breath error ($V_e - V_m$) in all subjects as a function of the average of $V_e$ and $V_m$ is illustrated in Fig. 7B. The mean difference between $V_e$ and $V_m$ (bias) was $0.009 \pm 0.046$ liters and was independent of $V_t$. Only two subjects, JS and SW, had values that spanned the entire confidence interval; the other subjects’ values were more tightly grouped, although some of them had a positive (QL) or negative (SW) bias.

Effect of head movements on volume estimation. In Fig. 8, the flow signal derived from tracheal breath sound amplitude (dotted line) is compared with measured flow (solid line) during head movements in a single subject. Figure 8A shows data obtained when the head was moved from side to side. Figure 8B illustrates up-and-down movements of the head.

Fig. 3. Relationship between flow and sound amplitude during quiet breathing in 8 subjects (identified by 2-letter initials). Note that, for a given flow, there is a wide between-subject variation in sound amplitude. In addition, at low flows, the sound signal does not exceed the background noise. Transfer function for conversion of sound to flow was calculated for inspiration and expiration separately over the quasi-linear portion of the relationship.
The error was not increased by full neck extension. The percent error of the VT estimates are shown in Table 1. They show that, with one exception, the error was considerably greater with neck rotation. No bias was introduced by VT size. Standard deviation (SD) was somewhat larger than that measured (0.046 liters) with the subject facing forward-facing, seated position were repeated after neck movement and postural changes in the six subjects, reproducibility of the estimation of VT could be estimated. The two volume signals remained comparable despite the intervening movement on the part of the patient. The mean difference between the two measurements was 0.001 liter, and the SD of the difference was 0.06 liter.

Reproducibility. Because measurements in the forward-facing, seated position were repeated after neck movement and postural changes in the six subjects, reproducibility of the estimation of VT could be estimated. The two volume signals remained comparable despite the intervening movement on the part of the patient. The mean difference between the two measurements was 0.001 liter, and the SD of the difference was 0.06 liter.

Ventilatory pattern. Table 3 gives values of VT, f, V_i, VT/Ttot, and Vv/Ti in the normal subjects and the patients. It is noteworthy that normal f averaged 19.3 breaths/min, VT averaged only 0.37 liter, and Vv/Ti averaged only 0.31 l/s. V_i was >2 SD greater than normal in the patients with unstable airways obstruction and in one of the asthmatic subjects. An increased Vv/Ti accounted for the increased V_i in these patients. One asthmatic patient had an unusually low f and a high VT.

In Fig. 10, the PDF of lnV_i (Fig. 10A) and VT (Fig. 10B) are shown. Dashed lines are normal subjects, and solid lines are patients. Patients tended to have larger V_i and VT. In Table 4, mean ± SD of V_i, VT, f, Vv/Ti, and VT/Ttot and the mean of their log transformed values are given along with the values of skewness and kurtosis for both the normal and log normal PDF. For each ventilatory parameter, the top row gives the normal values and the lower row the values in the patients. As judged by skewness and kurtosis, log normal PDF provided a better fit to the data with the sole exception of Vv/Ttot, in normal subjects.

**DISCUSSION**

Derivation of flow from sound. Continuous noninvasive measurement of flow and volume has both clinical and research applications. At present, the most widely used noninvasive technique for respiratory measurements is respiratory inductive plethysmography. However, with this technique, measured VT is affected by changes in posture and/or the shape of the chest wall (24, 35, 38, 39), limiting its usefulness in long-term monitoring. For this purpose, other noninvasive methods need to be developed. In this study, we have explored the feasibility of deriving ventilatory flow from...
Fig. 7. Comparison of estimated and measured tidal volumes (VT) during quiet breathing in 8 subjects. Each symbol represents data from a single subject. A: solid line represents the identity relationship; note that all data points fall close to the identity line, indicating a good correspondence between measured and estimated VT measurements. B: plot of the difference in volume measurement vs. means of the 2 measurements. Mean (±SD) difference was 0.009 ± 0.046 liter.

Fig. 6. Comparison of measured (solid lines) and estimated (dotted lines) flow signals in all subjects (identified by initials). Note that, in general, inspiratory flow is better estimated than expiratory flow.
The relationship between flow at the mouth and the amplitude of breath sounds has been under investigation for a number of years. Breath sounds obtained from microphones positioned at different places on the chest wall (3, 11, 17, 29, 32) and over the trachea (6, 9, 10, 33) have been examined. Some of these investigators have quantified the relationship between breath sound amplitude and tracheal flow by using mathematical functions. Banaszak et al. (3) described a linear relationship between flow and breath sound amplitude measured over the chest wall. A quadratic function was used by Shykoff et al. (32) and a power function by Gavriely and colleagues (10, 11). These functions have all provided a good fit to the experimental data; however, only in one paper (33) was there an attempt to use the derived mathematical relationship to obtain a flow signal from the sound signal. To our knowledge, Sofflet et al. (33) is the only paper that attempts to quantify tracheal breath sounds at flows of <0.5 l/s, a necessity if one wishes to record flows during quiet breathing. We confirmed their finding that there is a threshold flow that needs to be exceeded in order for sound to be present, and that even at flows greater than threshold the adverse SNR (Fig. 5) can present a major problem. Although the investigators recognized the threshold problem and the adverse SNR (evident from their data), they do not state how these major problems were solved.

Calculating respiratory flow from the breath sound signal is more complex than simply determining the relationship between the two signals. Because the breath sound amplitude at low flow rates does not exceed the background noise (Fig. 5), there exists a range of flows where no relationship between breath sound amplitude and flow can be obtained. Therefore, we derived the relationship between sound amplitude and flow only above the threshold value of sound that exceeded the background noise. The attempts to quantify the relationship between sound intensity and flow taken by others (3, 10, 11, 32) have not included this threshold and forces the relationship through the origin. This is clearly an oversimplification if one wishes to record the low flow rates used for quiet breathing.

Although above the threshold value the relationship between flow and sound might be better fit by a curve rather than a straight line, recent work (L.-G. Durand, unpublished observations) indicates that the scatter of the data relating flow to sound intensity (Fig. 3) does not allow curve fitting to provide a better estimate of VT than the linear approximation we used, which provided a reasonably accurate estimation of VT (Fig. 7). Thus we used a simple least-square linear regression to calculate the transfer function, and this was sufficient for a first attempt at deriving flow from breath sound amplitude.

During the period when flow was below the threshold value and the transfer function could not be used, we obtained flow values by linear interpolation from the lowest computed flow value to the end of either inspiration or expiration and then from these zero-flow points to the next recorded value of flow. Because the transitions between inspiration and expiration occupy only a small fraction of the total time of breathing and the flows are low during these periods, interpolation is quite accurate when the timing is distinguished correctly. It can be seen from Figs. 4 and 6 that Ve and Vm timing matched quite well despite the high threshold values used for the analysis. This was particularly true during the transition between inspiration and expiration due to the fast decelerations and accelerations during this part of the breathing cycle. Correspondence was not so good during the latter part of expiration and the transition to inspiration because some subjects had an end-expiratory pause. Although further refinements of the algorithm might enable us to better fit this part of the breathing cycle, accurate expiratory flows are not needed to obtain most ventilatory parameters of interest. Thus we calculated VT, f, V̇i, Ṫi, Ṫe, Ṫe/Ttot, and mean V̇e/Ṫi rate simply by measuring inspiratory flow alone as a function of time.

Sources of error. When we compared VT derived by integration of measured flow with the volume obtained by integration of estimated flow, in subjects in the seated position, we found a good correlation between the two values (Fig. 7A). The mean difference between Ve and Vm (0.001 liter) was insignificant, but individual breaths could vary by up to 0.1 liter (Fig. 7B).

Fig. 8. Estimated (dotted lines) and measured (solid lines) flows in different head positions in a single subject. A: head movement from left to right. B: up-and-down head movement. Horizontal lines indicate breaths in a given position. Note the good correspondence between estimated and measured flows in this subject even during the transition between different positions.
However, the error spanned the entire confidence interval in only two subjects (EB and JS). In subject SW, Ve was consistently higher than Vm, and, in JS, Ve was lower than Vm. A bias such as this can be characterized and accounted for when making measurements in an individual subject. As a result, the percent error in estimating Vₜ was <15% in all subjects when they were seated and facing forward (Table 1).

One explanation for errors in estimating Ve is breath-to-breath variation in the relationship between sound amplitude and flow (Fig. 3). Because the derived regression represents the best fit over a number of breaths, its application to individual breaths will result in differences compared with the volumes derived from integration of flow. Another source of variation could be filtering. Because most of the power in the tracheal sound signal is below 800 Hz (11), we chose 1,000 Hz as the low-pass cutoff for eliminating high-frequency noises and a sampling frequency of 3,000 Hz to exceed the Nyquist frequency. Tracheal sound fre-

Table 2. Effect of body position on error of Vₜ estimation

<table>
<thead>
<tr>
<th>Subject</th>
<th>Seated</th>
<th>Standing</th>
<th>Supine</th>
</tr>
</thead>
<tbody>
<tr>
<td>CK</td>
<td>0.09 ± 0.06</td>
<td>0.14 ± 0.11</td>
<td>0.25 ± 0.19</td>
</tr>
<tr>
<td>CL</td>
<td>0.15 ± 0.10</td>
<td>0.06 ± 0.03</td>
<td>0.20 ± 0.12</td>
</tr>
<tr>
<td>QL</td>
<td>0.09 ± 0.07</td>
<td>0.10 ± 0.07</td>
<td>0.38 ± 0.12</td>
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<tr>
<td>EB</td>
<td>0.08 ± 0.06</td>
<td>0.05 ± 0.04</td>
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<tr>
<td>SK</td>
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<td>0.15 ± 0.07</td>
<td>0.11 ± 0.08</td>
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<tr>
<td>SY</td>
<td>0.08 ± 0.08</td>
<td>0.09 ± 0.08</td>
<td>0.70 ± 0.26</td>
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</table>

Values are means ± SD.
Table 3. Ventilatory pattern

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<tr>
<th>Subject No.</th>
<th>VT, liter</th>
<th>VT l/min</th>
<th>Frequency, breaths/s</th>
<th>Ti/Ttot</th>
<th>VT/TI, l/s</th>
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</thead>
<tbody>
<tr>
<td>1</td>
<td>0.31 ± 0.06</td>
<td>6.9 ± 1.0</td>
<td>22.2 ± 2.3</td>
<td>0.39 ± 0.03</td>
<td>0.29 ± 0.04</td>
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<tr>
<td>2</td>
<td>0.49 ± 0.11</td>
<td>8.3 ± 1.5</td>
<td>17.3 ± 3.7</td>
<td>0.38 ± 0.06</td>
<td>0.36 ± 0.04</td>
</tr>
<tr>
<td>3</td>
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<td>7.3 ± 1.7</td>
<td>21.1 ± 4.3</td>
<td>0.45 ± 0.10</td>
<td>0.27 ± 0.04</td>
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<tr>
<td>4</td>
<td>0.43 ± 0.09</td>
<td>5.6 ± 1.3</td>
<td>13.1 ± 2.1</td>
<td>0.38 ± 0.07</td>
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<td>5</td>
<td>0.35 ± 0.07</td>
<td>7.9 ± 5.0</td>
<td>22.5 ± 4.6</td>
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<td>6</td>
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<td>7.4 ± 1.3</td>
<td>21.6 ± 3.2</td>
<td>0.34 ± 0.05</td>
<td>0.36 ± 0.03</td>
</tr>
<tr>
<td>7</td>
<td>0.32 ± 0.20</td>
<td>5.0 ± 1.2</td>
<td>17.4 ± 4.1</td>
<td>0.34 ± 0.07</td>
<td>0.25 ± 0.03</td>
</tr>
<tr>
<td>Mean</td>
<td>0.37 ± 0.065</td>
<td>6.9 ± 1.2</td>
<td>19.3 ± 3.5</td>
<td>0.37 ± 0.04</td>
<td>0.31 ± 0.06</td>
</tr>
</tbody>
</table>

**Normal Subjects**

<table>
<thead>
<tr>
<th>Subject No.</th>
<th>VT, liter</th>
<th>VT l/min</th>
<th>Frequency, breaths/s</th>
<th>Ti/Ttot</th>
<th>VT/TI, l/s</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.44 ± 0.12</td>
<td>15.2 ± 1.8*</td>
<td>22.1 ± 2.8</td>
<td>0.46 ± 0.08*</td>
<td>0.57 ± 0.09*</td>
</tr>
<tr>
<td>2</td>
<td>0.69 ± 0.11*</td>
<td>12.7 ± 2.6*</td>
<td>21.1 ± 2.9</td>
<td>0.28 ± 0.03†</td>
<td>0.74 ± 0.13*</td>
</tr>
<tr>
<td>3</td>
<td>0.61 ± 0.12*</td>
<td>9.9 ± 2.4*</td>
<td>27.7 ± 3.3*</td>
<td>0.33 ± 0.04</td>
<td>0.58 ± 0.11*</td>
</tr>
</tbody>
</table>

**Unstable Airway Obstruction**

<table>
<thead>
<tr>
<th>Subject No.</th>
<th>VT, liter</th>
<th>VT l/min</th>
<th>Frequency, breaths/s</th>
<th>Ti/Ttot</th>
<th>VT/TI, l/s</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.42 ± 0.19</td>
<td>9.0 ± 7.1</td>
<td>21.7 ± 6.7</td>
<td>0.31 ± 0.07</td>
<td>0.48 ± 0.34*</td>
</tr>
<tr>
<td>2</td>
<td>0.54 ± 0.17*</td>
<td>6.2 ± 1.5</td>
<td>17.1 ± 2.1†</td>
<td>0.35 ± 0.06</td>
<td>0.30 ± 0.07</td>
</tr>
<tr>
<td>3</td>
<td>0.50 ± 0.17</td>
<td>11.9 ± 2.7*</td>
<td>24.0 ± 4.3</td>
<td>0.34 ± 0.05</td>
<td>0.57 ± 0.10*</td>
</tr>
</tbody>
</table>

**Stable Asthmatic Patients**

Values are means ± SD. VT, minute ventilation; Ti/Ttot, duty cycle; VT/TI, inspiratory flow rate. *Value is greater than mean ± 2SD of normal subjects. †Value is less than mean − 2 SD of normal subjects.

Fig. 10. Log-normal (ln) probability density functions (PDF) of minute ventilation (VT; A) and VT (B) in normal subjects (dashed lines) and patients (solid lines).

ventilatory patterns can exceed 1,000 Hz (16), and these were not included in the sound envelope we measured. Nevertheless, our measurements of VT were reasonably accurate. In some subjects, estimation of zero flow points could pose a problem, and distinguishing inspiration from expiration might be difficult. Future refinements of the methodology should take these potential sources of error into account.

**Effect of head movement.** The estimated flow signal was not greatly affected by changing the head position from the seated to the standing position, which changes in volume would have resulted.

Moving from the seated to the standing position had little effect on VT estimation (Fig. 9; Table 2). This could be considered an advantage over inductance plethysmography, which changes cal-
vation with posture and changes in xiphipubic distance. We have not systematically measured the effect of thoracoabdominal configuration on the relationship between sound and flow in the trachea but see no a priori reason why it should change. Almost certainly, there was a significant change in xiphipubic distance between sitting and standing postures in at least some of our subjects. They were not instructed on how to sit during the recording period and merely chose a position that was comfortable. We subsequently measured changes in xiphipubic distance in normal subjects between seated and standing posture while maintaining a rigid upright spinal configuration and found changes between 5 and 8 cm.

The unacceptable error in changing posture from seated to supine indicates that the transfer function obtained in the seated position cannot be extrapolated to all positions. If this technology is to be used in the supine position, calibration of the sound signal needs to be done in that posture. This will be important because we have not yet shown that phonospirometry can be used supine, yet many of its clinical applications will require supine measurements. Other positions have yet to be investigated. Movement to the standing position did not affect the measurements; therefore, for longer measurements, a single calibration will probably suffice for standing and seated positions even with moderate head rotation flexion and extension.

Ventilatory pattern. In the preliminary results we obtained in normal subjects, we found a ventilatory pattern that differed in important ways from those summarized by Shephard (31). In contrast to the results shown in Table 3, he gives normal values taken from several studies of \( f \) that ranged from 10.1 to 15.8 breaths/min with a mean value of 13.2 breaths/min. Our value of 19.3 breaths/min is 46% higher than this. McCool and colleagues (21, 24, 25) measured the normal ventilatory pattern noninvasively from body surface measurements and found values for \( f \) that were only 10% higher than values obtained by spirometry. Our mean value for \( V_1 \) of 6.9 l/min is identical to the value reported by Shephard (31) and close to the value reported by Ashkanazi et al. (2) of 362 ml. They used a canopy system to make their measurements. Our value for \( V_1/Ti \) of 0.31 l/s is similar to the value of 0.298 l/s calculated from their data (2). They found that the primary effect of a mouthpiece and nose clip was to increase \( V_1 \) without change in \( Ti \) so that \( V_1/Ti \) increased to 400 ml/s.

It has been shown by several studies that a mouthpiece and nose clip alters breathing pattern. Although in some studies a decrease in \( f \) has been observed when breathing through a mouthpiece and nose clip (12, 13, 20, 27), this has not been found in other studies (2, 37). However, in reports where no significant effect on \( f \) was observed, the \( f \) on mouthpiece and nose clip was high, e.g., 19.1 breaths/min in Ashkanazi’s study (2). It seems clear that the normal \( f \) is considerably higher, whereas mean \( V_1/Ti \), \( V_1 \), and \( Ti \) are less than that frequently measured with a mouthpiece and nose clip.

Although the numbers are too small to draw any firm conclusions, Table 3 indicates that \( V_1 \) was abnormally increased in the three patients with unstable airway obstruction, and this was primarily due to an increase in mean \( V_1/Ti \). Two stable asthmatic patients also had an increased \( V_1/Ti \), one of whom had a high \( V_1 \). The third asthmatic patient had an abnormally low \( f \). The fact that all six patients had one or more abnormalities in ventilatory pattern suggests that phonospirometry will prove useful in detecting and quantifying abnormal ventilatory patterns in disease.

Ventilatory variability. We found considerable breath-to-breath variation in all ventilatory parameters. Nonrandom variations in ventilatory parameters are well described (see Ref. 4 for a review). Davis and Stagg (8a) found variations in \( V_1 \) and \( Ti \) with constant \( V_1/Ti \), whereas Ashkanazi et al. (2) found constant \( Ti \) and \( V_1/Tot \) with variations in \( V_1 \) and \( V_1/Ti \). We found variations in all these parameters. With the exception of \( Ti/Tot \), these were better described by log normal than by normal PDF.

Noninvasive measurements of ventilation. In this study, we have shown that, with proper calibration, \( V/Ti \) rate can be derived from a tracheal sound signal and that \( V \) can be estimated noninvasively to an accuracy within 15% of the measured volume signal. Because phonospirometry has the advantage of being
easy to calibrate and is insensitive to postural change from sitting to standing, it has advantages over other noninvasive methods of measuring ventilation, such as inductance plethysmography and magnetometry (21, 30, 34). Its tidal volume is equivalent to these techniques (20). It probably is not as accurate as optoelectronic plethysmography but is cheaper, easier to use, and less intrusive and does not require the subject’s position and posture to be restricted so that the thorax is continuously visualized by videocameras.

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REFERENCES