Noninvasive Estimate of Work of Breathing Due to the Endotracheal Tube

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Background: Although evidence suggests that secretions lining the inner wall of the endotracheal tube (ETT) often reduce its cross-sectional area, no data are available on the work of breathing as affected by the ETT. A noninvasive method is proposed for estimating the additional work of breathing necessary by the ETT in patients whose lungs are mechanically ventilated. This method (the acoustic-Blasius method) involves (1) determining the inner geometry of the ETT using the acoustic reflection method and (2) using these geometric data to solve the Blasius equation that characterizes the ETT pressure drop-flow relation.

Methods: To evaluate the acoustic-Blasius method in vitro, the authors computed the work of breathing due to the ETT in four healthy persons breathing through an ETT connected to a pressure-support device and in five tracheally intubated patients receiving mechanical assistance in the pressure-support mode. For the tracheally intubated patients, the reference value was the work calculated from the ETT pressure drop-measured between the two ends of the ETT using a pressure catheter.

Results: In the healthy participants and the tracheally intubated patients, there was close agreement between the inspiratory work per cycle values estimated by directly measuring the ETT pressure drop and calculated using the acoustic-Blasius method. The difference was consistently less than 0.08 joules (<10% of the reference value).

Conclusions: The data show that the acoustic-Blasius method allows noninvasive quantification of the ETT-related work of breathing in situ. (Key words: Work of breathing, endotracheal tube obstruction, noninvasive measurement of ETT patency.)

AVAILABLE data on endotracheal tube (ETT) pressure decreases during mechanical ventilation suggest that coating of the ETT wall by mucus reduces the cross-sectional area available for gas conduction. The resultant increase in apparent patient airway resistance may be interpreted mistakenly as indicating a respiratory abnormality rather than an isolated decrease in ETT patency. Decreased ETT patency may make mechanical ventilation hazardous and hinder spontaneous breathing while patients are being separated from mechanical ventilation. To our knowledge, there are no published data on measuring ETT-related work of breathing during mechanical ventilation.

Tracheal pressure has been measured by advancing a catheter to the distal end of the ETT, a procedure that requires precautions to ensure that the catheter is not obstructed and that its tip is placed in the distal part of the ETT. For these reasons, this procedure is not well adapted to routine use. In addition, the pressure catheter reduces the patent cross-sectional area of the ETT and alters the pressure-flow relation along the ETT. Tracheal pressure has also been estimated as the difference between the axial distance and the flow-dependent pressure drop across the ETT predicted from in vitro measurements. However, this approach cannot be used when ETT patency is altered by mucus deposition, because variations in the ETT cross-sectional area available for gas conduction are then responsible for unpredictable changes in the equations describing the ETT pressure-flow relation.

The acoustic reflection method was designed to determine the airflow cross-sectional area as a function of axial distance. Using this method, we detected, quantified, and located decreases in ETT area. In a separate study, we showed that the pressure-flow relation across the ETT can be described by the Blasius law.

Noninvasive measurement of ETT pressure drop-flow relation:

\[ \Delta P = \frac{n \cdot \Delta t}{\rho \cdot \Delta L} \]

where \( \Delta P \) is the pressure drop, \( n \) is the constant, \( \Delta t \) is the time interval, \( \rho \) the density of the gas, \( \Delta L \) the cross-sectional area, and \( \Delta t \) the time interval. The effect of mucus deposition on the wall of the ETT is

We propose a "locally" estimated pressure drop-flow relation:

\[ \Delta P = \frac{n \cdot \Delta t}{\rho \cdot \Delta L} \]

where \( \Delta P \) is the pressure drop, \( n \) is the constant, \( \Delta t \) is the time interval, \( \rho \) the density of the gas, \( \Delta L \) the cross-sectional area, and \( \Delta t \) the time interval. The effect of mucus deposition on the wall of the ETT is

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by directly measuring the pressure drop between the two ends of the ETT ($\Delta P_{\text{ETT}}$). The Blasius law allowed us to compute $\Delta P_{\text{ETT}}$ from the effective inner diameter of the ETT.

To estimate the additional work of breathing necessitated by the ETT during mechanical ventilation, we developed a noninvasive method (the acoustic-Blasius method) based on discrete application of the Blasius law to a set of ETT cross-sectional area values determined using the acoustic reflection method.

**Theory**

The pressure drop across a mucus-lined ETT is given by the Blasius resistance formula. This formula accurately relates inner ETT circular geometry (i.e., ETT diameter and length) to the pressure drop when the turbulent flow is fully developed and hydraulically smooth. Under these conditions

$$\Delta P = K \cdot \mu^{0.25} \cdot \rho^{0.75} \cdot L \cdot \left( \frac{1}{D^{1.75}} \right) \cdot \dot{V}^{1.75}$$

where $\Delta P$ is the pressure drop along the ETT, $L$ is the length of the ETT, $D$ is the diameter of the ETT, $K$ is a constant that depends on the shape of the tube cross section ($K = 0.24$ for circular tubes), $\mu$ is the viscosity, $\rho$ is the density, and $\dot{V}$ is the flow. In mucus-lined ETTs, the Blasius formula proved applicable despite modifications in the shape of both the perpendicular cross section and the longitudinal section, indicating that the sole effect of mucus deposition was to reduce the ETT diameter without altering the pressure drop law.

We postulated that the Blasius formula would remain "locally" applicable, and, that, consequently, the pressure drop $\Delta P$ could be expressed as the sum of discrete pressure decreases:

$$\Delta P = \sum_{n=1}^{n-1} \left[ K \cdot \mu^{0.25} \cdot \rho^{0.75} \cdot D \cdot \left( \frac{1}{(D_n \cdot \Delta L)^{1.75}} \right) \cdot \dot{V}^{1.75} \right]$$

$D(n \cdot \Delta L)$ is the actual ETT diameter at the distance $n \cdot \Delta L$. The cross-sectional area of the ETT, as a function of the distance along the longitudinal axis, was estimated using the acoustic method along the ETT with a spatial step increment of $\Delta L = 0.4$ cm.

The overall pressure drop across the ETT ($\Delta P_{\text{ETT}}$) (acoustic-Blasius method) was computed using the measured flow value and the actual ETT area profile (acoustic method) to solve equation 2.

Endotracheal tube work of breathing was then computed as $\int \Delta P_{\text{ETT}} \cdot \dot{V} \cdot dV$ over the inspiratory part of the breathing cycle. $\dot{V}$ was the volume derived from the flow measurement.

**Materials and Methods**

**Acoustic Reflection Method**

We used the two-microphone acoustic reflection apparatus developed by Louis and colleagues and recently used by van Sueil and coworkers to estimate endotracheal tube patency. This apparatus is considerably smaller than that used for the standard single-microphone technique. Briefly, a wave tube (0.9 cm internal diameter and 20 cm long) was prepared to accommodate two flush-mounted pressure transducers (piezoelectric pressure transducer 8510 B; Endevco, Le Pré-St-Gervais, France) and a horn driver (Motor 25 W; Bouyer, Montauban, France), as shown in figure 1. The distance between the transducers was 6.9 cm. The transducer nearest to the patient was 5.5 cm from the end of the wave tube, and the farthest was 6.5 cm from the horn driver. The wave tube opening was connected to the ETT via a standard connector. The horn driver was driven via a digital-to-analog converter by a computer-generated signal. Transducer outputs were fed to an analog-to-digital converter (14 bits) with a sampling period of 24 $\mu s$. The microcomputer derived the area-distance function from the digitized data using a computer program that was described previously (Hood Laboratories, Pembroke, MA). The complete wave forms recorded by the two microphones were used to compute the impulse response. Finally, the cross-sectional area of the ETT was obtained from the impulse response and the actual speed of sound.

This method accounts for possible ventilation-related variations in experimental conditions (gas composition, temperature, humidity, and pressure). The early part of the incident pressure signals was used to estimate the wave propagation speed relevant to the actual experimental conditions. We observed no significant variation in the acoustic results when measurements were taken during the respiratory cycle. However, we observed...
that the mechanical ventilation conditions were responsible for noise in the signals. Thus we introduced a new procedure into the algorithm to discard aberrant data caused by excessive noise during data acquisition. We took advantage of the fact that the initial segment of the computed area profile corresponds to the part of the wave tube located between the second microphone and the ETT connector. Because this segment has a constant, known area, we quantified the difference between the measured acoustic area profile and this reference profile. A threshold criterion allowed us to discard outlying measurements. Selected area profiles had a small coefficient of variation (less than 2%) for the five impulse responses (launched every 0.4 s) used to compute the cross-sectional area of the ETT.

**Work of Breathing, Pressure Drop, and Flow Measurements**

To estimate work of breathing, we asked healthy participants to breathe through an ETT connected to an intensive care pressure-support device (ARM 25; TAEMA, Antony, France) shown in figure 1 (setup A). The ETT was tightly fitted into a Plexiglas tube of 16 mm internal diameter connected to a mouthpiece. An additional tube of the same diameter as the ETT was placed between the ETT and the pneumotachograph to minimize the contribution of the entry flow effect to the ETT pressure drop. Pressure was measured using a differential pressure transducer (Validyne DP45 = 20 cm H₂O; Validyne, Northridge, CA). To measure actually the pressure drop across the ETT, the pressure transducer was connected to a previous hole drilled through the Plexiglas tube 1 cm away from the distal end of the ETT, and the other port was connected to a hole drilled in a standard ETT connector. Flow was measured using a pneumotachograph (Fleisch #2; Gould Electronique, Longjumeau, France) associated with a pressure transducer (Validyne DP45 = 3.5 cm H₂O).

Because of the abrupt change in area between the ETT and the Plexiglas tube, the measured pressure was not the pressure drop through the ETT (ΔPₑₑₑₑ). Indeed, the measured pressure drop included the convective-acceleration pressure change (Bernoulli effect) and the additional resistive pressure change (head loss) due to the abrupt expansion. Therefore, to obtain the ΔPₑₑₑₑ, we corrected the measured pressure as described by Loring and associates (see appendix 1).

To evaluate the acoustic-Blasius method in a clinical setting, we measured the ETT pressure drop and work of breathing in traumatically intubated patients whose lungs were mechanically ventilated and compared the results to those obtained using a reference method based on the introduction of a catheter into the ETT (fig. 1, setup B). A pressure transducer, connected to the mouth of a 2-mm catheter, was placed just below the mouth of intubated patients while breathing.

Flow was sampled at 128 Hz using an envelope detector to obtain the signals.

**Protocol**

Endotracheal intubation was performed on healthy volunteers who were asked to breathe through an ETT connected to a pressure-support device set at a constant rate of 10 breaths/min (Mallinckrodt, St Cloud, Minn) and a positive end-expiratory pressure of 5 cm H₂O. The ventilator was adjusted to keep tidal volume constant. The expiratory volume was compared to the inspiratory volume, which was measured using a flow-sensing strain gauge (model EX-00; Honeywell, Montreal, Que). Two liters of 1 l/min of compressed air was continuously supplied to the mouth of the healthy volunteer through a flexible tube that connected to the ETT.

**Endotracheal Intubation**

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(fig. 1, setup B). One port of the differential pressure transducer was connected to the proximal end of the ETT, as described for setup A, while the other port was connected to a pressure catheter. For this purpose, a 2-mm catheter with lateral holes but no distal opening was placed near the end of the ETT. This method allows measurement of the ETT pressure drop in vivo in patients whose lungs are mechanically ventilated.12–15

Flow and pressure signals were recorded and digitized at 128 Hz and were sampled by an analog/digital converter system (MP100; Biopac System, Goleta, CA). The signals were analyzed using a microcomputer.

Protocols

Endotracheal Tube Load during Normal Spontaneous Breathing. Four healthy participants were asked to breathe through an ETT connected to a pressure-support ventilator (setup A). Pressure support was set at three levels (0, 4, and 7 cm H2O). To simulate in vivo mucus-induced ETT obstruction, 8-mm ETTs (Mallinkrodt, Ireland) were coated with dry paint (Réline styrene J2; Julien, Marseille, France). The paint was introduced with a syringe into the distal part of the ETT. This produced an obstruction with a nonuniform longitudinal distribution. The distal obstruction was comparable to the obstruction that we observed in a previous study in intubated patients.12 Furthermore, as a result of the gravity effects when the paint was drying, the distal sections were no longer circular, because the paint accumulated in the dependent areas.

Two levels of obstruction were obtained using 5 and 6 ml of paint, respectively. To characterize these obstructed ETTs, we measured both the longitudinal acoustic area profile and the actual pressure decreases at different predetermined steady convective flows (in a range of 0.05–1.5 L/s).

Endotracheal Tube Load in Patients Receiving Pressure Support. In five patients in the medical intensive care unit of the Henri Mondor Hospital, pressure drop, flow, and acoustic parameters were measured using setup B. All five patients were being separated from ventilation and gave informed consent for the study. Table 1 shows their main characteristics. The patients were receiving ventilatory assistance in the pressure-support mode. The pressure-support level was increased by 3 cm H2O to compensate for the additional load prompted by the insertion of the wave tube into the ventilator circuit. An antibacterial filter was placed between the horn driver and the wave tube. Three to four respiratory cycles were analyzed in each participant.

Acoustic measurements were obtained after suction of mucous secretions. Because of the presence of the pressure catheter, the ETT pressure drop was computed as described in appendix 2.

Statistical Analysis

The results of the work of breathing obtained using the two methods (acoustic-Blasius method and catheter method) were compared by plotting the difference between the two values against the mean of the two values, as recommended by Bland and Altman.16

Results

Endotracheal Tube Load during Normal Spontaneous Breathing

Using steady flow conditions, we studied unused (fully patent) ET Ts and ETTs with the two levels of artificial obstruction with regard to ∆PET values obtained by direct measurement and by the acoustic-Blasius method (fig. 2). The pressure drop differences between the two methods remained low for all tested flows. Even for the highest flow and pressure-drop values, the absolute discrepancy between the measured and the acoustic-Blasius calculated pressure drop remained less than 0.5 cm H2O. The discrepancy was always less than 10% with turbulent flow conditions (i.e., Re ≈ 2,300 or flow >0.25 L/s).

Figure 3 shows typical ∆PET volume curves used to compute the ETT-related work of breathing under two different breathing conditions—spontaneous breathing and pressure support. We found good agreement between the results of the two methods (direct measurement and acoustic-Blasius method) in both the inspiratory and the expiratory limbs of the loops. The predicted ∆PET-volume curve showed small oscillations during the inspiratory phase.

In terms of the inspiratory work of breathing, differences between predicted and measured values were small. There were no significant differences between the two methods at any of the tested pressure-support levels (fig. 4). Mean differences ranged from 0.01 to 0.08 joules and tended to increase with the pressure-support level.
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Table 1. Main Patient Data

<table>
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<tr>
<th>Patient No.</th>
<th>PS (cmH2O)</th>
<th>PEEP (cmH2O)</th>
<th>ETT ID (mm)</th>
<th>Mechanical Ventilation (days)</th>
<th>Diagnosis</th>
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<td>0</td>
<td>8.5</td>
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<td>ARDS</td>
</tr>
<tr>
<td>2</td>
<td>25</td>
<td>3</td>
<td>7.5</td>
<td>6</td>
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<td>5</td>
<td>8</td>
<td>4</td>
<td>Liver transplantation</td>
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<tr>
<td>4</td>
<td>15</td>
<td>5</td>
<td>7.5</td>
<td>12</td>
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<tr>
<td>5</td>
<td>15</td>
<td>0</td>
<td>8.5</td>
<td>2</td>
<td>CABG</td>
</tr>
</tbody>
</table>

PS = pressure support level; PEEP = positive end-expiratory pressure level; ETT ID = endotracheal tube internal diameter; ARDS = adult respiratory distress syndrome; COPD = chronic obstructive pulmonary disease; CABG = coronary artery bypass graft.

Endotracheal Tube Load in Patients with Pressure Support

As shown in figure 5, measurements with the catheter method corresponded closely with estimates made using the acoustic-Blasius method. The absolute difference between the two methods was less than 0.05 joules and less than 10% of the measured inspiratory work.

In contrast, when equation 2 was solved using the ETT diameter given by the manufacturer instead of the measured acoustic profile, ETT-related work of breathing was underestimated by about 30% (fig. 5).

Discussion

Our results show that the pressure drop across the ETT can be estimated satisfactorily using a formula based on flow and on the ETT internal area profile as determined by an acoustic method. The pressure drop can then be used to estimate the work of breathing due to the ETT in spontaneously breathing or mechanically ventilated patients.

To predict the flow-related processes in tracheally intubated patients, several groups of researchers have developed empirical equations, such as the well-known Rohrer equation. Recently, Guttman and associates compared several mathematical models and found that the best fit of the pressure–flow relation was obtained using the equation \( \Delta P_{\text{ETT}} = b \cdot \dot{V}^a \). These authors validated their approach by comparing predicted tracheal pressures to tracheal pressures measured using the catheter technique in patients receiving mechanical ventilation. However, their procedure cannot be used to predict the \( \Delta P_{\text{ETT}} \)-flow relation when the cross-sectional area of the ETT changes, as a result of mucus deposition in vivo, for example. This is because the values of the coefficients \( a \) and \( b \) are usually determined in vitro for unused ETT and therefore do not account for a possible reduction in ETT patency.

In this study, we computed \( \Delta P_{\text{ETT}} \) using the Blasius formula, in which coefficient \( a \) is 1.75. Our results confirm that the Blasius formula provides an accurate estimation of the pressure–flow relation in ETTs, as demonstrated in a previous study. An advantage of the Blasius formula is that it accurately relates the inner ETT cross-sectional area to the pressure drop (equation 1) under in vivo conditions. The difference compared with the previous study is that the Blasius formula was applied to a series of segments, each of which had its own cross-sectional area, so that the pressure drop was calculated as the sum of several discrete pressure drops. In this summation procedure, it is implicitly assumed that the pressure drop in any segment of the tube is not influenced by entry effects. From the perspective of fluid mechanics, the reported formula is short

![Fig. 2. Pressure drop across the endotracheal tube (ETT) versus steady flow. The pressure drop was measured using setup B (see fig. 1; broken line) or computed using the acoustic-Blasius method (solid line). Pressure drop values were obtained for fully patent ETTs (a) and for ETTs with two levels of obstruction (b = lower level, and c = higher level). Standard error bars are too small to be seen in curves a and b.](image)

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Fig. 3. Volume versus pressure drop across the endotracheal tube (ETT) during a given respiratory cycle. Curves were obtained for a healthy participant using setup B described in figure 1, either by measuring the pressure drop (broken line) or by computing the pressure drop using the acoustic-Blasius method (solid line). Curves were plotted for ETTS with and without artificial obstruction (upper = fully patent ETTS; lower = obstructed ETTS) and in the presence and absence of mechanical assistance (left = no assistance; right = pressure support).

In fluid mechanics, the validity of this assumption is supported by the following arguments: (1) The entry length is shorter in a turbulent flow than in a laminar flow, and (2) the area decrease due to mucus deposition is smooth. Our experimental data confirmed the validity of this approach: measured and computed pressure decreases were very similar (fig. 2). At the lower end of the range of flows (V < 0.25 l/s), the Blasius formula is apparently not applicable. This value of 0.25 l/s corresponds to a Reynolds number of about 2,300, a critical value separating laminar flow from turbulent fully developed and hydraulically smooth flow. This value represents a theoretical limit for our method. In the range of low flows, the acoustic resistance underestimates the actual resistance. Nevertheless, the error on the pressure drop remains small (<0.2 cm H₂O). In practice, because (1) flow is less than 0.25 l/s, only during a short part of the inspiratory cycle, and (2) the resistance (and pressure drop) is much greater in the conditions of turbulent flow than of laminar flow, the error made on the computation of the work of breathing, when it is assumed that flow is turbulent throughout the inspiratory cycle, remains negligible. On the other hand, our method cannot be used without modifications in situations in which laminar flows are likely to be present within the physiologic range of flows, such as in neonates tracheally intubated with small ETTS. We found that even with the highest level of obstruction (at which the maximum local reduction in diameter—determined acoustically—was 23%) and the reduction in effective diameter—determined hydraulically—was 10%, as compared with the fully patent ETTS), the difference between predicted and measured pressure drop values was less than 10% at flows of 0.250 l/s or more. In a previous study, we found that at this level of ETT obstruction, emergency reintubation was required after a reduction of about 10% in the effective diameter. Thus, within this clinically relevant range of ETT obstruction, the acoustic-Blasius method remains applicable for predicting the ETT pressure drop.

We evaluated the acoustic-Blasius method in healthy participants breathing through an ET (fully patent or artificially obstructed) connected to a pressure-support device. There was close agreement between computed and measured pressure drops (figs. 3 and 4) under these conditions characterized by large flows and tidal volumes. The small oscillations in computed pressure drops seen in figure 3 were probably related to turbu-

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lent flow oscillations. In contrast, measured pressure drop-volume curves appeared smooth (fig. 3). This phenomenon can be ascribed to the frequency response of the pressure transducers, which act as a filter for high-frequency oscillations.17

The values of ETT-related work provided by the two methods were closely similar, the difference being less than 10% of the mean value. The slightly positive value of the mean of the difference between measured and computed values (fig. 4) is probably due to the fact that the entry effect and other nonlinear phenomena were disregarded. Another explanation for the slight discrepancy between theoretical and experimental results may

Fig. 4. Comparison of values for inspiratory endotracheal tube (ETT)-related work per cycle obtained using direct pressure measurement or the acoustic-Blasius method. As recommended by Bland and Altman,16 differences between the two values were plotted against the mean of the two values. The curves were obtained in healthy participants using setup A with three levels of pressure support (PS) (top: 0 cm H₂O, mean = 0.008 joules, SD = 0.017; middle: 4 cm H₂O, mean = 0.051 joules, SD = 0.039; bottom: 7 cm H₂O, mean = 0.080 joules, SD = 0.067). Each graph shows the data obtained for each of the four participants using ETTs with three levels of obstruction. The extra work necessitated by the ETT increased with the level of pressure support. This can be ascribed to the increase in flow associated with the increase in pressure support.

Fig. 5. Results obtained in tracheally intubated patients: comparison between the endotracheal tube (ETT)-related work of breathing measured by the catheter method and computed (predicted) using the acoustic-Blasius method, the Blasius formula, and the theoretical EIT-Blasius method, given by the manufacturer. Data were obtained from three or four cycles in each of the five patients. (Upper) Inspiratory work per cycle computed using the acoustic-Blasius method versus inspiratory work measured using the catheter method (correlation coefficient = 0.97). The solid line is the identity line. (Lower) As recommended by Bland and Altman,16 differences between the results of the two methods were plotted versus the mean of the two methods (mean = 0.011 joules, SD = 0.027).

be that, to compute D[n · ∆L] from the acoustic area, we implicitly assumed a pure circular geometry, which obviously did not match reality during the in vitro or the in vivo tests. Nevertheless, these results suggest that the acoustic-Blasius method is appropriate for quantifying the in vitro additional work per breath related to an obstructed ETT under dynamic flow conditions.

Reliable acoustic data were obtained in tracheally intubated patients receiving mechanical assistance with pres-
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In patients whose lungs were mechanically ventilated, we found that values of ETT-related work of breathing were similar to the acoustic-Blasius method and to the direct-measurement method (fig. 5). In contrast, ETT-related work was underestimated when the theoretical ETT diameter given by the manufacturer was used in the Blasius formula (fig. 5), confirming the value of the acoustic method. This underestimation of ETT work was due to overestimation of the actual ETT area. Because the pressure decrease given by the Blasius formula depends on 1/D², slight overestimation of the ETT area results in marked underestimation of the pressure decrease in the ETT and of the ETT-related work of breathing. For example, in patient 3, whose ETT was considered by the clinical team to be free of mucus deposition, the difference between the area given by the manufacturer and the area estimated using the acoustic method was about 10% (5% for the ETT diameter) (fig. 6). As a result, during the inspiratory cycle, the ETT pressure drop was underestimated by 4 cm H₂O and ETT-related work of breathing by 27% (fig. 6). These substantial quantitative errors would be dramatically increased in patients with ETT obstructions similar to those reported by van Surell and coworkers² (36% area reduction).

Knowledge of the ETT-related work of breathing may be clinically important when ventilation is in the pressure-support mode. Indeed, part of the ventilatory assistance is used to overcome the resistive load due to the ETT. Underestimation of the ETT pressure drop may lead to errors in the mechanical analysis of the respiratory status of the patient. Accurate estimation of the ETT-related work of breathing may be especially important when patients are separated from the mechni-
nal ventilation process, because the level of pressure support is among the parameters used to determine whether such separation is appropriate.

We found that our acoustic-Blasius method provides an accurate estimation of the additional work of breathing related to the ETT. Because quantification of the mechanical consequences of partial ETT obstruction may be critical both for monitoring artificial ventilation and for making therapeutic decisions in intensive care units, the acoustic-Blasius method should be studied in more patients.

The authors thank Dominique Touchard for drawing the figures.

Appendix 1: Measuring Endotracheal Tube Pressure Drop

The pressure drop, $\Delta P$, measured using setup A in figure 1, is the sum of (1) the pressure drop in the ETT, $\Delta P_{\text{ETT}}$; (2) the resistive pressure drop (head loss), $\Delta P_{\text{res}}$ due to abrupt expansion during inspiratory flow (or abrupt contraction during expiratory flow); and (3) the convective accelerative pressure change, $\Delta P_{\text{acc}}$, due to acceleration produced by changes in the cross-sectional area of the tube: $\Delta P = \Delta P_{\text{ETT}} + \Delta P_{\text{res}} + \Delta P_{\text{acc}}$.

By convention, the pressure differences $\Delta P_{\text{ETT}}$, $\Delta P_{\text{res}}$, $\Delta P_{\text{acc}}$, and $\Delta P$ are the pressures in section 1 minus the pressures in section 2 (setup A, fig. 1). Furthermore, the flow is positive when it travels from section 1 to section 2, as during inspiration.

$\Delta P_{\text{acc}}$ is obtained from the Bernoulli equation, as follows:

$$\Delta P_{\text{acc}} = \frac{1}{2} \rho \left( \frac{V^2}{A_2} - \frac{V^2}{A_1} \right)$$

where $A_1$ and $A_2$ are the cross-sectional areas of sections 1 and 2, respectively (see fig. 1, setup A).

$\Delta P_{\text{res}}$ is given by

$$\Delta P_{\text{res}} = \frac{1}{2} \rho \left( \frac{V^2}{A_2} \right)$$

for an abrupt expansion and a positive flow, and by

$$\Delta P_{\text{res}} = -\frac{1}{2} \rho \left( \frac{V^2}{A_1} \right)$$

for an abrupt contraction and a negative flow. $\epsilon$ is a constant that depends on the $A_1/A_2$ ratio and on the flow regime. When flow is turbulent and $A_1/A_2 = 1/4$, as in this experimental setup, $\epsilon$ is close to 0.3. Thus we have

$$\Delta P_{\text{ETT}} = \Delta P + \frac{1}{2} \rho \left( \frac{V^2}{A_2} \right)$$

during expiratory flow.

$$\Delta P_{\text{ETT}} = \Delta P + \frac{1}{2} \rho \left( \frac{V^2}{A_1} \right)$$

during inspiratory flow.

Appendix 2: Endotracheal Tube Pressure–Flow Relation with a Catheter

The nondimensional form of the Blasius resistance formula stays valid when the catheter diameter, $d$, is considerably smaller than the ETT diameter, $D$.

$$\frac{\Delta P}{\rho \mu \frac{V}{A_1}} = \frac{3.166 Re^{-0.25}}{D_d}$$

with $Re = \frac{\rho V A}{\mu}$.

$D_d$ is the hydraulic diameter (here $D - d$), $Re$ is the Reynolds number defined with the hydraulic diameter, and $A$ is the cross-sectional area of the ETT. The pressure decrease is given by

$$\Delta P = K \rho \mu^{0.75} \mu^{0.25} \frac{V^2}{D_d}$$

with $K = 0.24$.

In this case, equation 2, which allows us to derive the pressure drop from the acoustic data, was replaced by

$$\Delta P = \sum_{n=1}^{m} K \rho \mu^{0.75} \mu^{0.25} \frac{V^2}{D_d}$$

References

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