Dual assessment of airway area profile and respiratory input impedance from a single transient wave

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1Unité de Physiopathologie et Thérapeutique Respiratoires INSERM U492, Service de Réanimation Médicale, Hôpital Henri Mondor, AP-HP, 94010 Créteil; 2Service de Réanimation Néonatàle, Hôpital Cochin Port-Royal, AP-HP, 75679 Paris; and 3Service d’Explorations Fonctionnelles, Hôpital Raymond Poincaré, AP-HP, 92380 Garches, France
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Louis, Bruno, Redouane Fodil, Samir Jaber, Jérôme Pigeot, Pierre-Henri Jarreau, Frédéric Lofaso, and Daniel Isabey. Dual assessment of airway area profile and respiratory input impedance from a single transient wave. J Appl Physiol 90: 630–637, 2001.—This report concerns the inference of geometric and mechanical airway characteristics based on information derived from a single transient planar wave recorded at the airway opening. We describe a new method to simultaneously measure upper airway area and respiratory input impedance by performing dual analysis of a single pressure wave. The algorithms required to reconstruct airway dimensions and mechanical characteristics were developed, implemented, and tested with reference to known physical models. Our method appears suitable to estimate, even under severe intensive care unit conditions, the respiratory system frequency response (above 10 Hz) in intubated patients and the patency of the endotracheal tube used to connect the patients to the ventilator.

airway geometry; mechanical response; forced oscillations; acoustic reflection

NONINVASIVE EVALUATION of airway geometry and the mechanical properties of the human respiratory system has remained a constant challenge for clinicians and engineers, especially in spontaneously breathing subjects. Two distinct methods, based on analysis of small, time-varying external forces, are currently available to perform this type of geometric/mechanical assessment: 1) the acoustic reflection method, initially proposed in humans by Fredberg et al. (10), which provides the airway geometric properties, and 2) the forced-oscillation method, first described by Dubois et al. (6), which provides the mechanical properties of the entire respiratory system. In present applications, these two methods use two distinct types of excitation signals, i.e., a transient pulse for the acoustic reflection method and a steady state of periodic oscillations or pseudorandom noise input for the forced-oscillation method, making concomitant assessment of geometric/mechanical properties difficult. An obvious improvement would therefore consist of simultaneous application of the two external forcing methods to combine the previously demonstrated clinical benefits of each into a single method. Briefly, the acoustic reflection method provides a spatial representation of proximal airways, whereas the forced-oscillation method provides the mechanical properties of the entire respiratory system, including distal airways, but without providing precise information about their spatial distribution. Furthermore, the acoustic reflection method not only allows oral, nasal, and tracheal airway assessments (16), but recent miniaturization of the recording apparatus has also allowed this method to be applied in unconscious, intubated intensive care patients (18, 24). The forced-oscillation method has been predominantly used in nonintubated patients rather than intubated subjects.

To simultaneously apply the acoustic reflection method and the forced-oscillation method, we propose dual analysis of a single transient wave to allow concomitant assessment of the geometric and mechanical properties of the respiratory system. This new approach, together with the associated theory derived from wave propagation equations, described below, constitutes a new method dedicated to dual assessment of upper airway area and respiratory input impedance (called AAI).

METHODS AND PROTOCOL

Methods

Area profile. The acoustic reflection method, which gives the longitudinal cross-sectional area profile along the airway, has already been described, and its limits have been thoroughly defined for the various living applications (3, 7, 10, 14, 17, 24). This method is based on analysis of a planar acoustic wave propagating in a rigid duct connected to the airway. Briefly, oscillatory pressure and oscillatory flow rate in a duct can both be described by the sum of two waves propagating in opposite directions with the same wave speed (20)

\[ P(\omega, x) = P_2(\omega, x) + P_1(\omega, x) \]  \hspace{1cm} (1)

\[ V(\omega, x) = \frac{1}{Z_c(\omega, x)} \cdot [P_1(\omega, x) - P_2(\omega, x)] \]  \hspace{1cm} (2)

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where $P$ is the pressure, $V$ is the flow, $\omega$ is the pulsation, and $x$ is the spatial coordinate. $P_i$, the incident wave, is the pressure wave propagating in the positive $x$ direction, whereas $P_r$, the reflected wave, is the pressure wave propagating in the negative $x$ direction. $Z_c$, called characteristic impedance, reflects the relationship between pressure and flow waves and tends toward $(\rho c^2/\lambda)(x)$ with increasing values of pulsation where $c$ is the wave speed, $A$ is the cross-sectional area, and $\rho$ is the gas density. Using this equation system, written in the time domain, Ware and Aki (26) showed that $A(x)$ can be inferred from the value of the impulse response under a set of restrictive conditions. This impulse response, $H$, is defined in the frequency domain as

$$P_i(x, \omega) = H(\omega, x) \cdot P(\omega, x) \quad (3)$$

Previous studies have shown that measurement of the impulse response can be used to infer the area-vs.-distance function from the mouth to the end of the trachea (10), from the nostril to the sinuses (15), or from the ventilator connector to the distal end of an endotracheal tube (ETT) (24).

Impedance. The impulse response can be used to obtain the input impedance at the entry of the airways according to the transitory forced-oscillation method described by Fredberg et al. (10). The impedance, $Z$, is classically defined in the frequency domain as the ratio between pressure and flow. $Z$ can be computed from Eqs. 1–3

$$Z(\omega, x = 0) = \frac{P(\omega, 0)}{V(\omega, 0)} = \frac{Z_c(\omega, 0) \cdot 1 + H(\omega, 0)}{1 - H(\omega, 0)} \quad (4)$$

If a wave tube is connected to the respiratory system, the value of $H$ at the end of the wave tube can be used to compute the respiratory impedance by Eq. 4. Equation 4 does not require any assumption concerning the system measured, which simply corresponds to the load impedance at the end of the wave tube. Respiratory impedance, computed from the impulse response and wave-tube characteristics, therefore, corresponds to the overall result of the entire respiratory system, including distal airways, tissues, and chest wall properties.

The standard transmission-line formalism (1, 4, 20a) can be used to compute the characteristic impedance and propagation constant of a tube. These parameters can be used to obtain the impedance of a tube of finite length, and the relationship between the input impedances can be computed at two sites of a tube with a constant section separated by a distance $\Delta L$ (at $x$ and $x + \Delta L$).

$$Z(\omega, x) = Z_c(\omega, x) = \frac{Z(\omega, x + \Delta L) + Z_c(\omega, x) \cdot \tgh[\beta(\omega, x) \cdot \Delta L]}{Z_c(\omega, x) + Z(\omega, x + \Delta L) \cdot \tgh[\beta(\omega, x) \cdot \Delta L]}$$

$$Z(\omega, x + \Delta L) = Z_c(\omega, x) \cdot \frac{Z(\omega, x) \cdot \tgh[\beta(\omega, x) \cdot \Delta L] - Z(\omega, x)}{Z_c(\omega, x) \cdot \tgh[\beta(\omega, x) \cdot \Delta L] - Z_c(\omega, x)} \quad (5)$$

where $\beta$ is the propagation constant of the tube and $\tgh$ designates the hyperbolic tangents. Several authors (1, 4) have shown that $Z_c$ and $\beta$ mainly depend on the Womersley parameter, $\alpha$, and the Prandtl number, $Pr$: $\alpha = d^2/2(\omega \nu)^{0.5}$, where $d$ is the diameter and $\nu$ is the kinematic viscosity. For a given viscosity, $\alpha$ can be considered to be a function of $A$ and $\omega$: $\alpha = A(\omega/\nu)^{0.5}$. $Pr$ is a thermodynamic parameter used to characterize the compressibility of the gas used. For a given gas, i.e., for given $Pr$ and $\nu$, Eq. 5 can therefore be used to construct an incremental procedure on the spatial coordinates to compute the impedance at any point of a circular rigid-wall duct with a variable longitudinal cross-sectional area

$$Z(\omega, x = 0) \quad \text{Eq. 5} \quad Z(\omega, \Delta L)$$

$$\frac{Z_c(\alpha, A(x = 0))}{\alpha, A(\Delta L)} \quad \text{Eq. 5} \quad Z(\omega, x = n \Delta L) \quad (6)$$

where $n$ is an integer. The input impedance at a given point, $x_n$, appears to be a function of two main variables: the input impedance at the origin and the area profile between this origin and the point $x_n$.

$Z_c$ and $\beta$ were computed by assuming an isothermal tube wall, as proposed by Brown (4) and Benade (1). The impulse response was measured by the two-microphone method developed by Louis et al. (18). With this method, the impedance responses are inferred from measurement of the pressure at two distinct loci in a wave tube ($x = -L$ and $x = 0$). By comparison with the classical one-microphone method (10), this method allows a marked reduction of the setup size because it does not require explicit separation between the incident and reflected waves. In this study, we used a deconvolution algorithm in the frequency domain.

Setup. The experimental device used to measure area and impedance consisted of a wave tube with two microphones (piezoresistive pressure transducers 8510-B; Endevco France, Le Pré-Saint-Gervais, France) and a horn driver (Fig. 1) tightly connected to the proximal airway opening. A transient acoustic wave was generated by the horn driver driven by a personal computer via a digital-to-analog converter. The resulting pressure in the wave tube was recorded via the microphones for a period of ~0.2 s. The peak-to-peak amplitude of the oscillating pressure was ~1 cmH₂O. The frequency content of the pressure included the range 10–4,500 Hz. Microphone outputs were fed to an analog-to-digital converter (14 bits, 24-μs sampling period, E. Benson Hood Laboratories, Pembroke, MA) and recorded on a personal computer. The discretized pressures (8,192 points/channel) were converted into the frequency domain and analyzed as indicated above to infer impulse response, area, and input impedance. The choice of recording procedure, i.e., sampling frequency and number of points recorded, introduced a spatial increment of ~0.4 cm for $A(x)$ and a frequency increment of ~5 Hz for $Z(V)$. On each day of measurement,
the two transducers were self-calibrated, as indicated in the appendix.

Protocol

AAI was tested by measuring various physical models made from cylindrical Plexiglas tubes of various diameters (models 1, 2, and 3). The complete characteristics of the models used are given in Table 1. The measured impedance was compared with a theoretical model given by the analytical solution of the laminar oscillatory compressible flow (1, 4), considered to be the reference method. We ignored entry effects in the analytical solution, even when abrupt diameter variations occurred. Two wave tubes were used: a 1.9-cm-diameter tube for pulmonary application at the mouth and a 0.8-cm-diameter tube for application in ventilated subjects via an ETT. In both cases, the two microphones were separated by a distance of 6.9 cm.

The ability of AAI to estimate impedance beyond a given distance of ETT, as indicated by Eq. 6, was tested by measuring the impedance of a known model (model 4) with the 0.8-cm-diameter wave tube. This model consisted of an 8-mm ETT connected to a circular tube with a constant diameter. Impedance was measured either from the site of the second microphone, as indicated in Eq. 6, or immediately beyond the distal extremity of the ETT, by using Eqs. 5 and 6.

Moreover, because the frequency response of these models was very different from that of the physiological model, at least in the low- to medium-frequency range (e.g., the real part is much lower than the imaginary part, in contrast with airway impedance), we constructed a physical model with a frequency response resembling the physiological response. We used a Plexiglas tube (length 10 cm and inner diameter 1.55 cm) filled with a parallel network of small capillaries made of undulated aluminum foil connected to the inner volume of a rigid spherical chamber (inner volume, 1,725 cm$^3$, inner diameter $\sim 15$ cm). This resulted in an increase in the real part of the impedance at the lower frequency range (<100 Hz) associated with a negative imaginary part at the lowest frequency. The distal end of a 7-mm ETT was introduced into the Plexiglas tube. The ETT cuff was used to obtain a good seal between the ETT and Plexiglas tube. ETT produced into the Plexiglas tube. The ETT cuff was used to lower frequency. The distal end of a 7-mm ETT was introduced into the wave tube, as indicated in Eq. 6, or immediately beyond the distal extremity of the ETT.

Results

Figure 2 presents the results obtained with the AAI method in two mechanical models (models 1 and 2 in Table 1) using the 1.9-cm-diameter wave tube. The inner diameters of these tubes (wave tube and models) roughly correspond to oral applications in adults. The areas inferred with the AAI method were in agreement with real values (Fig. 2A), i.e., a first 2.8-cm$^2$ tube followed by a second 2-cm$^2$ tube for model 1 and a 2.8-cm$^2$ tube closed at its end for model 2. The impedance values measured over the entire frequency range (10–2,000 Hz) were close to the reference values for both the real and imaginary parts (Fig. 2B) for model 1, whereas a marked discrepancy was observed between reference and measured values with model 2 at the lowest frequency.

Table 1. Physical models

<table>
<thead>
<tr>
<th>Model No.</th>
<th>Tube 1</th>
<th>Tube 2</th>
<th>Boundary Conditions</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Diameter 1.9 cm</td>
<td>Diameter 1.6 cm</td>
<td>Open to the atmosphere</td>
</tr>
<tr>
<td>2</td>
<td>Length 5.4 cm</td>
<td>Length 114 cm</td>
<td>Without</td>
</tr>
<tr>
<td>3</td>
<td>Diameter 1.9 cm</td>
<td>Diameter 0.92 cm</td>
<td>Open to the atmosphere</td>
</tr>
<tr>
<td>4</td>
<td>Length 24.5 cm</td>
<td>Length 59.8 cm</td>
<td>Diameter 1.9 cm</td>
</tr>
<tr>
<td>More relevant physiological model</td>
<td>Diameter 0.81 cm</td>
<td>Diameter 1.74 cm</td>
<td>Open to the atmosphere</td>
</tr>
<tr>
<td></td>
<td>Length 8 cm</td>
<td>Length 10 cm</td>
<td>With a network of small capillaries</td>
</tr>
<tr>
<td></td>
<td>8-mm ETT</td>
<td>Length 17.4 cm</td>
<td>Connected to a 1,725-cm$^3$ rigid chamber</td>
</tr>
<tr>
<td></td>
<td>Length 31 cm</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>7-mm ETT</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Each physical model consists of a tube (tube 1) connected to another tube (tube 2). The second extremity of the second tube has specific boundary conditions. When a second tube is not used, the boundary conditions describe the second extremity of the first tube. ETT designates endotracheal tube.

Man diagnostic applications, e.g., evaluation of mechanical nonhomogeneity and effect of drugs such as histamine (2, 19, 21, 23, 27), were performed by using more or less implicitly lumped-parameter models, we limited the frequency range of interest up to 90 Hz for the physiological model, at which point the validity of lumped parameters starts to be questionable. Above this frequency, the length of the respiratory system may no longer be negligible in relation to the pressure wavelength, and this model may no longer be more physiologically relevant than other rigid-wall long-tube models used in this study (models 1, 2, 3, and 4 in Table 1). The results obtained with the physiologically relevant physical model were compared with a “reference” method that consisted of directly measuring the model without the ETT in between the wave tube and the model.

The AAI method was also tested in two intubated patients in the Henri Mondor Hospital intensive care unit. Measurements were performed in the context of a protocol approved by the Ethics Committee of the Societe de Reanimation de Langue Francaise (French Society of Intensive Care Medicine). Informed consent was obtained from the patients or their closest relatives. One patient had a history of chronic obstructive pulmonary disease (COPD), whereas the other was intubated with no respiratory disease (no-RD). The patients were briefly disconnected from the ventilator and breathed spontaneously during the measurement, as routinely performed during an endotracheal suctioning procedure. One measurement consisted of 10 pulses (total duration $\sim 2$ s). Each pulse was analyzed separately in terms of area vs. distance and impedance vs. frequency.
The results obtained with the AAI method (in model 3) with the 0.8-cm-diameter wave tube, i.e., the wave tube dedicated to the ETT application, were very similar to those observed with the 1.9-cm wave tube. Areas and impedance measured with the AAI method were still in agreement with theoretical values (Fig. 3).

To describe the effect of the locus used to estimate the impedance for ETT applications, we compared the impedance measured at the end of an 8-mm ETT in model 4 (Fig. 4). Only impedances measured from the distal end of the ETT, i.e., by Eqs. 5 and 6, were in agreement with the theoretical values of these models over the entire frequency range tested. In particular, the impedances for which a relative maximum in terms of impedance was predicted by the reference method were correctly observed when the impedance was measured at the distal end of the ETT. In contrast, the impedance measured at the second microphone, located close to the proximal end of the ETT, $Z(0)$, was very different from the theoretical value.

Evaluation of the AAI method performed in the physiologically relevant physical model (see Protocol, above) is presented in Fig. 5 for a 7-mm ETT. Longitudinal area profiles were closely correlated with the expected area profiles (Fig. 5A). Without obstruction inside the 7-mm ETT, the area of the ETT was found to be $0.4$ cm$^2$. When full cylinders with cross-sectional areas of 0.07 and 0.2 cm$^2$ were successively introduced inside the 7-mm ETT to simulate ETT obstruction, the resulting partial obstructions were detected, quantified, and located (Fig. 5A). The 0.07-cm$^2$ cylinder induced a moderate reduction in ETT area (20%), whereas the 0.2-cm$^2$ cylinder induced a major reduc-

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**Fig. 2.** Area profile and input impedance of models 1 and 2 measured with the 1.9-cm-diameter wave tube. A: area vs. distance for these 2 models. *Model 1* (solid line) is a 5.4-cm-long tube with a 2.8-cm$^2$ constant area connected to a 114-cm-long tube with a 2.0-cm$^2$ constant area open to the atmosphere. *Model 2* (dashed line) is a 24.5-cm-long tube with a 2.8-cm$^2$ constant area closed with a rigid stopper. B: real and imaginary parts of impedance ($Z$) of *model 1* measured at $x = 0$, $Z(0)$ ( ), compared with the analytical solution (1, 4) of the oscillatory compressible flow (solid line) considered to be the reference method. C: same as B but for *model 2*.

**Fig. 3.** Area profile and input impedance of model 3 measured with the 0.81-cm-diameter wave tube. A: area vs. distance of model 3. The model is an 8-cm-long tube with a 0.5-cm$^2$ constant area connected to a 59.8-cm-long tube with a 0.7-cm$^2$ constant area open to the atmosphere. B: real and imaginary parts of impedance of model 3 measured at $x = 0$, $Z(0)$ ( ), compared with the analytical solution (1, 4) of the oscillatory compressible flow (solid line) considered to be the reference method.
tion (50%). For all obstruction conditions, i.e., no, moderate, or major obstruction, the impedance measured at the second microphone \(Z(0)\); Fig. 5B] clearly differed from the impedance of the physical model obtained by the “reference” method, i.e., the AAI method without an ETT between the wave tube and the model (see Protocol). In contrast, for no obstruction and moderate obstruction, the impedance measured at the distal end of the ETT, \(Z(\text{ETT end})\), was fairly close to that measured with the reference method (Fig. 5B). Under conditions of major obstruction, the real part of impedance measured at the distal end of the ETT was slightly overestimated compared with the reference method, whereas the imaginary part was in agreement with the reference method.

Evaluation of the AAI method in two intensive care patients (COPD and no-RD; see Protocol) showed that the ETT of the COPD patient was not obstructed, whereas the ETT of the no-RD patient was slightly obstructed near its distal end (Fig. 6A). In these patients, estimation of impedance at the end of the ETT, \(Z(\text{ETT end})\), compared with estimation at the second microphone, \(Z(0)\), reduced the frequency dependency of the real part of impedance, at least in the clinically relevant frequency range (Fig. 6B).

DISCUSSION

It is generally accepted that oscillation methods constitute powerful tools to explore the respiratory system. Together with modeling and parameter estimations, these methods have been shown to be valuable to detect and interpret respiratory system abnormalities associated with disease. However, clinical applications, especially in the intensive care unit, have remained limited for various reasons, including the considerable measuring time compared with the respiratory time required for the forced-oscillation method. The acoustic reflection method setup has also been too cumbersome for intensive care unit applications, before the minia-

Fig. 4. Evaluation of the dual assessment of upper airway area and respiratory input impedance (AAI) method to estimate impedance beyond a given distance of ETT. A: area vs. distance of an 8-mm ETT connected to a 17.4-cm-long tube with a 2.8-cm² constant area open to the atmosphere. B: real and imaginary parts of impedance, estimated by the AAI method at the proximal end of the ETT, \(Z(0)\) (+), and at the distal end of the ETT, \(Z(\text{ETT end})\)(○), were compared with the analytical solution of the 17.4-cm-long tube (1, 4) (solid line) considered to be the reference method.

Fig. 5. Effect of obstruction on estimation of impedance at the distal end of the ETT by the AAI method. A: area vs. distance of a 19.5-cm-long 7-mm ETT without obstruction, with moderate obstruction and with major obstruction. Moderate obstruction was simulated with a cylinder 6.8 cm long and 0.07 cm² in cross-sectional area placed 4.5 cm from the distal end of the ETT, whereas major obstruction was simulated with another cylinder 3 cm long and 0.20 cm² in cross-sectional area placed 5 cm from the distal end of the ETT. B: real and imaginary parts of the more physiological model estimated at the proximal end of the ETT, \(Z(0)\) (solid symbols), and at the distal end of the ETT, \(Z(\text{ETT end})\) (open symbols), under all conditions of obstruction, i.e., no obstruction (diamonds), minor obstruction (triangles), and major obstruction (circles). These impedance measurements are compared with the results of the AAI method without ETT between the wave tube and the model considered to be the reference method (solid line).
The present study demonstrates that a single transient wave presents a sufficient frequency content to assess both airway geometry and respiratory input impedance from analysis of its reflection. Our approach can be considered to be a further improvement of both oscillation methods, in which the two methods are combined in a single method, AAI, with 1) simultaneous measurement of area profile and respiratory impedance and 2) a miniaturized setup that has already been shown to be valuable for intensive care applications.

**Physical implications.** Measurements of input impedance inferred by acoustic reflection can be used to estimate, by the incremental procedure described in Eq. 6, the input impedance from the distal end of the ETT, provided that the ETT retains a circular shape (Figs. 4 and 5). This property remains true when the shape of the ETT is moderately altered, i.e., when the area variation is <20\% (Fig. 5). In the case of a marked alteration of shape, such as the major ETT obstruction simulated in this study (Fig. 5), the diameter derived from the area measurement can no longer be used to compute a Witzig-Womersley number, \( \alpha \), representing the ratio between tube radius and the viscous boundary layer (1, 8). Altering the circular shape of a tube tends to modify the ratio between tube radius and viscous boundary layer, i.e., \( \alpha \). For high values of \( \alpha (\alpha > 5) \), the viscous loss, i.e., the tube resistance, increases with \( \alpha \). In contrast, inertia and compliance of the gas tend to become constant with increasing values of \( \alpha \). Consequently, altering the circular shape of a tube tends mainly to modify tube resistance, without severely altering either inertia or compressibility of the gas. This may explain why impedance measured at the end of an ETT with a major obstruction was overestimated the real part of impedance, at least at low frequencies, whereas the imaginary part was correctly estimated (Fig. 5).

**Physiological implications.** Estimating respiratory impedance from the distal end of the ETT by the AAI
DUAL ASSESSMENT OF AIRWAY AREA AND INPUT IMPEDANCE

method is clinically important because it provides a representative index of the patient’s respiratory system, which is not affected by the geometric characteristics (diameter and length) of the ETT that vary from one patient to another. The AAI method provides respiratory system impedance from the end of the ETT only if the ETT is not obstructed or only moderately obstructed (≥20%). However, for measurements of the ETT area profile, the AAI method easily detects a major obstruction, allowing the concomitant altered respiratory system impedance measurement to be disregarded. It should be noted that the major obstruction tested in Fig. 5 is a clinically unacceptable situation because, during a trial of weaning, this degree of ETT obstruction would dramatically increase the patient’s work of breathing with a risk of failure of the weaning procedure (5, 24, 25). Similarly, during controlled mechanical ventilation, a major ETT obstruction remains clinically unacceptable because it increases the risk of total ETT obstruction with possibly fatal consequences.

In both of these clinical situations, a therapeutic procedure (suction maneuver, ETT extubation and reintubation with a new ETT, etc.) would be performed to restore ETT patency and normal ventilatory conditions.

Elimination of the influence of the ETT on the real part of impedance, as performed on the data obtained with the two patients (Fig. 6B), facilitates interpretation of the patient’s respiratory resistance. This resistance was −3.5 cmH2O·l−1·s for the no-RD patient and −14 cmH2O·l−1·s for the COPD patient. These values are in the high ranges of values usually obtained in nonintubated patients with and without obstructive lung disease, respectively (21). We also observed that estimation of impedance from the end of the ETT increased the first resonance frequency, i.e., the frequency beyond which the imaginary part of the impedance becomes positive (Fig. 6B). Computing impedance from the end of the ETT obviously eliminates the influence of the ETT inertia component. For example, in the case of the no-RD patient, the imaginary part estimated from x = 0 appeared to be essentially inertial (Fig. 6B), but this was no longer true when the influence of the ETT was eliminated by estimating impedance from the distal end of the ETT.

Although most of the routine forced-oscillation studies allowing human diagnostic applications, such as evaluation of mechanical nonhomogeneity and bronchial reactivity of drugs like histamine, were conducted in the low- to medium-frequency range (below 40 Hz), it would be clinically useful to develop a method able to measure impedance in the medium- to high-frequency range. For instance, measurement of the medium- to high-frequency impedance (8–2,048 Hz) combined with an anatomic structural modeling approach can be used to assess the change in distribution of airway diameters after bronchoconstriction (12).

In the present application of the AAI method, the lowest frequency accepted for correct impedance measurement was ~10 Hz. This frequency is slightly above the lowest frequency (3–6 Hz) studied by the conventional forced-oscillation method to investigate obstructive lung disease (21, 22) in living human beings. It is clear that this minor difference in the frequency limit will not really change the results of evaluation of the patient’s status based on an index such as the mean value of real impedance over a frequency range. Figure 6 shows that the two obstructive and nonobstructive patients presented two distinct behaviors. In contrast, the 10 Hz limit may constitute a problem when data analysis includes the first resonance frequency, as in the case (12) of determination of tissue compliance according to the classical Dubois six-element method, because, in healthy subjects, this first resonance frequency is generally lower than 10 Hz. Because estimation of impedance from the distal end of the ETT tended to increase the first resonance frequency (Fig. 6B), this problem will essentially concern nonintubated subjects. We also found that the low-frequency limit could be decreased to 5 Hz (data not shown) by increasing the frequency range of the excitation signal and by doubling the acquisition time. Nevertheless, the capacity of the AAI method, whose main advantage is to allow measurements in intensive care patients, to explore the low-frequency range (<10 Hz), remains questionable because of the difficulties of measuring very high-impedance values at low frequencies and because of the very large number of data that need to be recorded in this setting.

A method to estimate impedance from the distal end of an ETT has already been proposed by Habib et al. (11). This method requires preliminary calibration for each ETT based on separate measurements of each ETT loaded by two distinct and known impedance conditions (ETT closed and ETT open). This method implicitly assumes that ETT shape remains unchanged between calibration and measurement, although the ETT shape may become altered by mucus deposition or if the patient bites the ETT. Because of these assumptions associated with this preliminary calibration, Habib’s method is difficult to apply in the intensive care unit. By comparison, our proposed AAI method does not require such assumptions and therefore appears to be more suitable for intensive care unit applications. Moreover, the AAI method allows assessment of ETT area in parallel with respiratory system impedance.

In summary, we have developed a method allowing simultaneous assessment of 1) upper airway geometry, i.e., from the mouth to the carina or to the distal end of the ETT in intubated patients, and 2) respiratory input impedance from 10 to 2,000 Hz, from a single signal recorded at the entry of the airway. This AAI method is completely noninvasive and easy to use in the intensive care unit.

APPENDIX

Calibration. The wave tube was connected to a tube, called the calibration tube, of length l and with the same diameter, d, as the wave tube. The input impedance of this calibration tube, Zin, can be estimated by the analytical solution of
compressible flow for viscous gases (1, 4)

\[ Z_L = Z_c \cdot \left[ \frac{Z_{rad} + Z_c \cdot \tgh(\beta L')}{[Z_c + Z_{rad} \cdot \tgh(\beta L')]^2} \right] \]

where \( Z_{rad} \) is the radiation impedance for an unflanged long tube (13)

\[ \rho \cdot \omega^2/4 \pi c \cdot j (2.44 \cdot \rho/d) \cdot \omega/2\pi \]

(see Methods for definitions of variables). The ratio between the pressure at two points of a wave tube, separated by a distance \( L \), is theoretically given by

\[ \frac{[P(\omega, -L)/P(\omega, 0)]_{\text{theory}}}{\Delta P_{\text{theory}}} = \frac{Z_c \cdot \sinh(\beta L) + \cosh(\beta L)}{Z_L} \]

The ratio between the theoretical pressure, \( \Delta P_{\text{theory}} \), and the corresponding measured quantity, \( \Delta P_{\text{measured}} \), provides a calibration for the two transducers at each frequency

\[ \Delta P_{\text{corrected}} = \Delta P_{\text{measured}} \cdot \left( \frac{\Delta P_{\text{theory}}}{\Delta P_{\text{measured}}} \right) \quad \text{and/or} \]

\[ P(\omega, -L)_{\text{corrected}} = P(\omega, -L)_{\text{measured}} \cdot \left( \frac{\Delta P_{\text{theory}}}{\Delta P_{\text{measured}}} \right) \]

The calibration procedure takes into account differences in frequency response between the two channels due to various reasons such as the electronic circuit and pressure tap geometry.

REFERENCES


