Effects of posture and bronchoconstriction on low-frequency input and transfer impedances in humans

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Dellaca, Raffaele L., Lauren D. Black, Haytham Atileh, Antonio Pedotti, and Kenneth R. Lutchen. Effects of posture and bronchoconstriction on low-frequency input and transfer impedances in humans. J Appl Physiol 97: 109–118, 2004. First published February 13, 2004; 10.1152/japplphysiol.00721.2003.—We simultaneously evaluated the mechanical response of the total respiratory system, lung, and chest wall to changes in posture and to bronchoconstriction. We synthesized the optimal ventilation waveform (OVW) approach, which simultaneously provides ventilation and multifrequency forcing, with optoelectronic plethysmography (OEP) to measure chest wall flow globally and locally. We applied an OVW containing six frequencies from 0.156 to 4.6 Hz to the mouth of six healthy men in the seated and supine positions, before and after methacholine challenge. We measured mouth, esophageal, and transpulmonary pressures, airway flow by pneumotachometry, and total chest wall, pulmonary rib cage, and abdominal volumes by OEP. We computed total respiratory, lung, and chest wall input impedances and the total and regional transfer impedances (Ztr). These data were appropriately sensitive to changes in posture, showing added resistance in supine vs. seated position. The Ztr were also highly sensitive to lung constriction, more so than input impedance, as the former is minimally distorted by shunting of flow into alveolar gas compression and air walls. Local impedances show that, during bronchoconstriction and at typical breathing frequencies, the contribution of the abdomen becomes amplified relative to the rib cage. A similar redistribution occurs when passing from seated to supine. These data suggest that the OEP-OVW approach for measuring Ztr could noninvasively track important lung and respiratory conditions, even in subjects who cannot cooperate. Applications might range from routine evaluation of airway hyperreactivity in asthmatic subjects to critical conditions in the supine position during mechanical ventilation.

Special Article

THEORY

The main objective of the present study is to combine the OVW method with OEP to study how posture and bronchoconstriction influence mechanical properties of the total respiratory system, the lung, and the chest wall (inclusive of its individual compartments) for frequencies surrounding breathing. Our data were acquired by forcing at the mouth and measuring the flow both at the mouth and at the chest wall surface (as the time derivative of the total and compartmental Vcw) (3, 15). In this way, we can compare input (Zin) and transfer impedances (Ztr) (see Theory) over a frequency range not previously explored by both simultaneously. Thus another objective of this study was to establish if the local impedance and total Ztr (which do not require an esophageal balloon) at lower frequencies are sensitive to important changes in lung mechanics that occur with changes in posture or airway provocation.

Theory. To first approximation, and particularly at lower frequencies, the respiratory system can be conceptualized as a two-port T-network system (Fig. 1A) in which the airway

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Subjects and protocols. Six healthy nonsmoking men (age 30 ± 8.8 yr, height 179.6 ± 5.8 cm, weight 84.8 ± 8.5 kg) were studied following the same procedure in both the seated and the supine positions on 2 different days in random order. Our institutional research committees approved the study, with informed consent from each subject required. Before the experiment started, an esophageal balloon was passed transnasally and positioned as described in Measurements. The measurement of the impedances requires the relaxation of all of the respiratory muscles. This condition was achieved after a training period of 20–40 min, with the subject connected to the servo-controlled ventilator. During this period, the subjects learned how to completely relax their respiratory muscles and to be ventilated by the OVW system for at least 45–60 s. Pao was continuously monitored, and the training was concluded when Pao showed no breathing efforts made by the subject during the application of the OVW.

Impedances were measured by asking the subjects to relax passively to functional residual capacity and place their mouth around the OVW mouthpiece while an operator was firmly supporting their cheeks. The ventilator was started such that the first OVW inspiration began from functional residual capacity. The waveform was then presented for 40–100 s. This procedure was repeated three or four times, and the data were stored for subsequent analysis. After this, subjects were administered increasing concentrations of aerosolized methacoline (MCh) solution delivered by a nebulizer system (New Rosenthal Dosimeter, Pulmonary Data Services). Three minutes after each MCh administration, the OVW measurement was performed. We used MCh concentrations of 0.01, 0.1, 1, 10, and 25 mg/mL. The MCh administrations were continued until the maximum dose was reached or until the R at the lowest frequency increased to more than three times the baseline value. We used the lowest frequency R because it is most sensitive to all phenomena associated with bronchoconstriction, including airway and tissue R and heterogeneities. To allow the MCh to wash out, at least 48 h passed before the experiment was repeated in the other posture.

Forced oscillations. The experimental setup is summarized in Fig. 2. The forcing signal was generated by a personal computer connected to an analog-to-digital-to-analog board (DAQ 6068E, National Instruments, Austin, TX) and sent to the system used to generate the flow forced signal. This system was made by a servo-controlled linear motor (Inomag model 7315–1, Goleta, CA) connected to a piston-cylinder arrangement, as described in Ref. 30. The flow-forcing signal was made by adding six sinusoids, whose frequencies were chosen according to the nonsum-nondifference order 3 criteria (41). The sinusoids were chosen with frequencies of 0.156, 0.39, 0.86, 1.48, 2.42, and 4.61 Hz, with the lowest frequency amplitude being three-to fourfold bigger than the amplitude of the other components. The amplitude of the resulting signal was adjusted to provide a tidal volume of ~600–800 ml.

Measurements. Pao was measured by a piezoresistive pressure transducer (model SCX05, SenSym, Milpitas, CA) connected to the mouthpiece. Vao was measured with a pneumotachograph (Fleisch no. 2) connected to a pressure transducer (model TDI05, ±5 cmH2O; SCIREQ, Montreal, QC). Pes was measured by a pressure transducer (model SCX05, SenSym, Milpitas, CA) connected to a 10-cm-long thin latex balloon positioned in the lower one-third of the esophagus and filled with 0.4 ml of air. Transpulmonary pressure was obtained by subtracting Pes from Pao. The position of the esophageal balloon was considered correct if the transpulmonary pressure remained almost unchanged during gentle efforts against a shutter connected to the

![Diagram A](image)

**Table 1. Definition of the impedances**

<table>
<thead>
<tr>
<th>Element</th>
<th>Input Impedance</th>
<th>Transfer Impedances</th>
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<tbody>
<tr>
<td>Total respiratory system</td>
<td>Zin = Pao / Vao</td>
<td>Ztr = Pao / Vbs Zrc,p = Pao / Vrc,p Zab = Pao / Vab</td>
</tr>
<tr>
<td>Lung</td>
<td>ZL = P / Vao</td>
<td></td>
</tr>
<tr>
<td>Chest wall</td>
<td>Zcw = Pes / Vbs</td>
<td></td>
</tr>
</tbody>
</table>

Z, impedance; P, pressure; V, flow; in, input; ao, airway opening; t, lung; cw, chest wall; es, esophageal; bs, body surface; tr, transfer; rc,p, pulmonary rib cage; ab, abdominal.
mouthpiece and closed at end expiration of a tidal breath (10), and 2) if the cardiogenic oscillations were minimized. Special care was used for the balloon positioning in the supine position (11). All of the signals from the transducers were amplified, low-pass filtered at 40 Hz (model SC-24, SCIREQ), and digitized at 200 Hz by the same analog-to-digital-digital-to-analog board used to generate the forcing signal. Vcw was measured plethysmographically by OEP, a recently introduced technique extensively validated in several conditions and postures (2, 4, 5, 13, 17, 24). Briefly, several (89 for the seated position and 68 for the supine) reflective markers were placed over the thorax from the clavicles to the anterior superior iliac spines along predefined vertical and horizontal lines. The position of the markers is described in Ref. 13 for the seated position and in Ref. 15 for the supine. The three-dimensional position of each marker was measured by an automatic motion analyzer (ELITE System, BTS, Milano, Italy) at a frequency of 100 Hz by using four special video cameras (two in front and two behind the subject for the seated position, two on the left-hand side and two on the right-hand side above the subject for the supine, Fig. 2). Vcw was finally determined by approximating chest wall surface by triangles connecting the markers and computing the volume enclosed by all of these triangles. The use of a subset of the markers allowed us to measure separately the volume of the three chest wall compartments defined in the Theory section, subdividing Vcw into the RCp volume, the RCA volume, and the abdominal volume. These data permit estimation of local impedances (Table 1). This technique was proven to be reliable when used to measure total and compartmental tidal volume, as well as to estimate the total respiratory system Ztr, at least up to 24 Hz (3). Moreover, because it is based on the measurement of volumes, it is not affected by integration drift, and, therefore, it can be used to track breath-by-breath end-expiratory lung volume changes (16) and thus measure the possible total and compartmental dynamic hyperinflation induced by bronchoconstriction.

Data analysis: Impedances Z have been estimated as follows (32):

\[ Z = \frac{G_{pf}}{G_{ff}} \quad (1) \]

where \( G_{pf} \) is the cross spectrum between pressure and flow signals, and \( G_{ff} \) is the auto-spectrum of the flow signal.

The estimates of \( G_{pf} \) and \( G_{ff} \) have been obtained by averaging different periodograms computed by applying the fast Fourier transform to several data segments. To reduce the picket-fence effect, we resampled the acquired data at a new sampling frequency of 40 Hz. The resampling procedure was preceded by an anti-aliasing low-pass digital filter compensated for the filter’s delay. Because our forcing signal period was 12.8 s, one OVW period after the resampling was 512 samples long. In this way, all frequencies of the OVW signal perfectly matched the discrete frequency points for which the fast Fourier transform computes the exact behavior.

The signal segments were selected manually by an operator to discard those with esophageal spasms or breathing efforts. The Ztr were computed by measuring pressure and volume and by multiplying the Fourier transform of the volume signal by \( \omega_0 \), where \( \omega \) is the imaginary unit, \( \omega_0 \) is equal to \( 2\pi f \), and \( f \) is the frequency (in Hz). This procedure is equivalent to differentiating the volume signal in the time domain to obtain the flow signal (3).

The coherence functions of the impedances were computed as described in Ref. 3. In the present study, we considered only values of impedances with a coherence value >0.90.

The transfer functions of the entire population of subjects were finally obtained by averaging the real and the imaginary part of the transfer function values from all of the subjects at all of the frequencies. If the value of the impedance was discarded for one frequency because its coherence was below the threshold (likely due to cardiogenic oscillations in Pes or to low signal-to-noise ratio in volume signal), it was estimated by linear interpolation by using the adjacent values.

In this study, all of the impedances are presented in terms of R, i.e., the real part, and E, which were obtained by multiplying the imaginary part by \(-2\pi f\), where \( f \) is the frequency in Hz.

Statistical analysis. Values are presented as means ± SE, if not otherwise specified. A nonparametric Wilcoxon test determined statistical significance of changes in all of the considered transfer functions with changes in posture or after bronchoconstriction. The test was performed for both R and E at each forcing frequency. Significance was taken as \( P < 0.05 \).

RESULTS

Individual values of Zin, Ztr, and ZL measured in the seated position are shown in Fig. 3 as examples of original data before all transfer functions of the population were pooled. The variability of the data between different subjects is similar for Zin and Ztr, even if the two impedances were obtained by using different flow measurement techniques (pneumotachography for Zin and OEP for Ztr).

Total Ztr at low frequencies via OEP method. In Fig. 4, Ztr, Zin, ZL, and chest wall Zin (Zcw) spectra are shown as R and
E. Data are shown for seated and supine positions. In the seated position, Z\textsubscript{L} appears as reported previously, with lung R showing a slight frequency-dependent drop and lung E a small increase to 1 Hz followed by a decrease (22, 23). The Z\textsubscript{cw} shows significantly more frequency dependence in both R and E. The Z\textsubscript{tr} and Z\textsubscript{in} are very similar at all frequencies, with the R from Z\textsubscript{in} being slightly smaller than that from Z\textsubscript{tr} at every frequency, and the Z\textsubscript{in} E being smaller at only the higher frequencies. In the supine position, impedances become amplified, and there is a greater difference between Z\textsubscript{in} and Z\textsubscript{tr}, particularly for R. The E of Z\textsubscript{tr} at the highest frequency showed a significant reduction and an increase of intersubject variability as we were close to the resonant frequency [that is, lower for Z\textsubscript{tr} than for Z\textsubscript{in} (29)]. Despite the fact that Z\textsubscript{L} was measured by pneumotachography, whereas Z\textsubscript{cw} was measured by OEP, the sum of Z\textsubscript{L} and Z\textsubscript{cw} was very close to the Z\textsubscript{tr}. In fact, the average percent difference between the sum (series combination) of Z\textsubscript{L} and Z\textsubscript{cw} and the Z\textsubscript{tr} on the six frequencies was $4.3 \pm 4.6$ (SD) % for seated and $1.3 \pm 10.1$ (SD) % for supine (Fig. 4).

**Effect of posture on impedance at low frequencies.** The real part of Z\textsubscript{L} (lung R) slightly increases in the supine position (Fig. 4), probably due to the effect of lower lung volume (7, 34). When comparing seated to supine, a large increase in frequency dependence and in intersubject variability can be observed in the chest wall R. With posture change, no significant changes were observed in lung E, whereas a small change in frequency dependence was present for Z\textsubscript{cw}.

To further examine the impact of posture change, we can partition Z\textsubscript{tr} into its local impedance components (Fig. 5). Here we do not show the RC\textsubscript{a} compartment, as its contribution to V\textsubscript{cw} was found to be negligible compared with the RC\textsubscript{p} and Ab. A large increase in the RC\textsubscript{p} R can be observed when passing from the seated to the supine position. Conversely, the R of the abdominal compartment showed a small reduction in the supine position compared with the seated one. Therefore, while in the seated position the two compartments showed comparable R values, in the supine position they offered very different R values. Due to the fact that the two compartments are in parallel (they are submitted to the same pressure, Pes), and because in the supine position the RC\textsubscript{p} impedance had a very high baseline value, in this posture the total Z\textsubscript{tr} R is determined mostly by the abdominal compartment, showing only small differences compared with the seated posture.

From seated to supine, the E of the RC\textsubscript{p} showed a significant increase of both mean level and frequency dependence. Abdominal E decreased in both mean level and frequency dependence. For this reason, in the supine position, most of the flow...
Fig. 4. Resistance (top) and elastance (bottom) of ZL (●), chest wall impedance (Zcw; ○), Zin (○), and Ztr (▲) in control conditions. Data are presented for the seated (left) and the supine (right) positions. Values are means ± SE. *P < 0.05, seated vs. supine.

Fig. 5. Total Ztr (▲), pulmonary rib cage (Ztr,rcp; ●), and abdominal pathway impedances (Ztr,ab; ○) expressed as resistances (top) and elastances (bottom) for the seated (left) and the supine (right) position. Values are means ± SE. *P < 0.05, seated vs. supine.
induced by the ventilator is displaced by the Ab. Thus, whereas the two compartments showed large changes, they were in opposite directions and, as occurred for R values, only slight differences were observed in total Ztr.

**Ztr vs. ZL during bronchoconstriction.** In Fig. 6, the effects of bronchoconstriction on ZL, Zcw, and Ztr are shown for the supine position. We show supine only because the effect of the MCh was greatest in this position. Most of the effects of MCh on Ztr are due to changes in lung mechanics. In fact, Zt dramatically increased both in mean level and in frequency dependence. As reported previously (26), these changes are highly consistent with acute heterogeneous constriction occurring throughout the peripheral airway tree, inclusive of airway closures or near closures. Hence the noninvasive measure of Ztr via the OEP method is highly sensitive to heterogeneous constriction in the lung.

The chest wall R showed no significant changes after MCh. Slight differences are present in chest wall E, likely due to the dynamic hyperinflation induced by bronchoconstriction. By using OEP, it was possible to measure dynamic hyperinflation as differences between the absolute Vcw values at end expiration before and after the challenge. The results are shown in Table 2, and they indicate that our subjects were hyperinflated at end expiration before connecting to the OVW system by 0.435 ± 0.230 liters when seated and by 0.431 ± 0.080 liters when in the supine position. While the amount of hyperinflation was similar in the two postures, the pattern of the distribution of the hyperinflation was significantly different, with the Ab contributing 21.6% to the total volume changes in the seated and 50.6% in the supine position.

**Total Zin vs. total Ztr.** Figure 7 shows the comparison for Zin and Ztr total respiratory impedances measured simultaneously, in the seated and supine position and for control vs. heterogeneous constriction, as induced by MCh. Postchallenge, both Zin and Ztr show substantial increases in the levels and frequency dependence of R and E, with a larger effect occurring in the supine position. The difference between seated and supine was larger postconstriction. Postconstriction, the increase in Ztr was greater than that of Zin in the supine position. The increases in E were more similar for Zin and Ztr.

**DISCUSSION**

There has been a resurgence of interest in measuring the frequency dependence of lung and/or respiratory dynamic properties for frequencies surrounding breathing. Modeling and experimental studies confirm that these data reflect 1) the degree of airway constriction, 2) the relative contributions of tissue and airways to breathing mechanics (42), 3) the pattern,

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**Table 2. Compartmental changes in end-expiratory chest wall volumes after bronchoconstriction**

<table>
<thead>
<tr>
<th></th>
<th>ΔEEVrc,p</th>
<th>ΔEEVab</th>
<th>ΔEEVcw</th>
</tr>
</thead>
<tbody>
<tr>
<td>Seated</td>
<td>Mean</td>
<td>SD</td>
<td>Mean</td>
</tr>
<tr>
<td></td>
<td>0.252±0.215</td>
<td>0.094±0.280</td>
<td>0.435±0.514</td>
</tr>
<tr>
<td>Supine</td>
<td>0.099±0.169</td>
<td>0.218±0.162</td>
<td></td>
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</tbody>
</table>

Values are means ± SD in liters. Δ, Compartmental change; EE, end-expiratory; Vrc,p, Vab, Vcw: pulmonary rib cage, abdominal, and chest wall volume, respectively.
or heterogeneity, of lung disease (18, 26), and 4) the local and global contributions of the chest wall (8), all in the context of spontaneous breathing. Forcing methods have been proposed and used to measure lung properties from 0.1 to 4 Hz in awake humans, but they require an esophageal balloon. If applied to the total respiratory system, they result in a measure of Zin that is entirely noninvasive. However, during disease conditions that elevate ZL, the Zin could be overly sensitive to the shunt impedance of the upper airways (14, 36, 37). An alternative measure that is also entirely noninvasive is Ztr, and, in principle, this method is less sensitive to upper airway shunting. The reason is that the flow measured at the airway opening (used to determine Zin) is not equal to flow at the chest wall surface (used to determine Ztr). In fact, as shown in Fig. 1, flow at the airway opening is equal to the flow at the thoracoabdominal surface added to the flow shunted by gas compression or upper airway walls. Because Ztr uses chest wall flow, shunt pathways affect it less.

However, no previous method could acquire Ztr reliably at low frequencies in human subjects. This is because the measurement of Ztr requires either the linear excitation of pressure surrounding the entire chest wall or a direct measure of chest wall flow. The former is not practical, especially with loudspeaker technology.

In this study, we defined a new method to measure low-frequency mechanical properties of the respiratory system by combining the optimal ventilation waveform technique, used to provide both ventilation and forcing signal, with the OEP, used to measure chest wall flow. There are several advantages to this approach of measuring Ztr: 1) it does not require the use of a plethysmographic chamber, and it can be applied in different postures and conditions; 2) it allows the separation between the different rib cage and abdominal pathways; 3) it allows the simultaneous assessment of Zin and Ztr; and 4) it allows the breath-by-breath tracking of lung volume changes and chest wall configuration through the direct measurement of total and compartmental end-expiratory Vcw. Moreover, if combined with Pes measurement, this technique allows the partitioning of lung and chest wall mechanical properties by the measurement of the impedance of each single component (see Theory).

Comparison between Zin and Ztr. The first important step was to validate that the OEP-based method provided a reliable approach to estimating Ztr. Previous studies by our group have validated the ability of OEP to measure volume changes and changes in end-expiratory lung volumes (4, 13), as well as its ability to measure Ztr during single sinusoidal forcing at higher frequencies (4–20 Hz) (3, 15). Validation of OEP at the lower frequencies used in this study is evident from two aspects of our data. First, despite the fact that ZL was measured by pneumotachography whereas Zcw was measured by OEP, the sum of ZL and Zcw was very close to Ztr (see RESULTS). Second, several other studies have presented data on low-frequency Zin (6, 20, 30), and they reported Zin values that are similar to ours. To our knowledge, there are no other studies presenting low-frequency Ztr measured by forcing at the mouth. However, we expected that Zin and Ztr would differ only because of a distinct influence of shunt impedances on Zin. Specifically, when forcing at the mouth, upper airway shunt and alveolar gas compression will tend to reduce Zin but not impact Ztr (27, 36). This effect should be least evident during the control
sitting conditions and more evident as $Z_{\text{L}}$ increases. Indeed, we found that $Z_{\text{in}}$ and $Z_{\text{tr}}$ were very similar for control and seated position measurements and that moving from seated to supine caused the $R$ of $Z_{\text{tr}}$ to increase slightly compared with that of $Z_{\text{in}}$ (Fig. 4). After bronchoconstriction, the differences between $Z_{\text{in}}$ and $Z_{\text{tr}}$ became more substantial (Fig. 7), and this is consistent with an increased influence of shunt on $Z_{\text{in}}$ but not on $Z_{\text{tr}}$.

In this study, we partitioned chest wall from lung mechanical properties by estimating pleural pressure swings by the esophageal balloon technique in both the seated and the supine positions. It has been shown that, in the supine posture, Pes swings represent an accurate estimation of pleural pressure swings (11). This is supported by the very similar values obtained for low-frequency lung $E$ in the two postures ($6.1 \pm 0.8$ cmH$_2$O/l for the seated position and $6.7 \pm 0.7$ cmH$_2$O/l for the supine).

In summary, $Z_{\text{tr}}$ and $Z_{\text{cw}}$ measured by using OEP are consistent with $Z_{\text{in}}$ and $Z_{\text{L}}$ simultaneously measured by pneumotachometry, showing that the combination of OEP and OVW provides a reliable method for measuring low-frequency impedances.

Analysis of respiratory impedances and the impact of posture. In the seated position, lung $R$ (Fig. 4) is consistent with a relatively homogeneous mechanical system with some slight frequency dependence due to viscoelastic lung tissues (23). Most of the frequency dependence of $Z_{\text{tr}}$ at low frequency is, therefore, due to the chest wall tissues. Our data show that, around breathing frequencies, the chest wall contributes $\sim 62\%$ to the total $Z_{\text{tr}}$ $R$ in the seated position, whereas, at 4.6 Hz, $Z_{\text{cw}}$ contributes only $24\%$ of the total $Z_{\text{tr}}$. Similar results for $Z_{\text{cw}}$ were found by Barnas et al. (8), measuring chest wall flow by a partial body plethysmograph, and by Hantos et al. (20), estimating $Z_{\text{cw}}$ from $Z_{\text{in}}$ after correcting for upper airways and thoracic gas shunt.

The change in posture slightly increased the mean level and frequency dependence of $Z_{\text{L}}$ (Fig. 4). These changes are compatible with the differences in lung volumes and airway diameters anticipated for supine vs. seated positions. The increase in frequency dependence of $R$ and $E$ is consistent with the supine position, enhancing airway heterogeneities even at baseline. A reduced lung volume causes reduction of the elastic recoil forces acting on the airways, which is combined with the change of the dependent regions of the lung (due to the different direction of the gravitational forces), inducing a different degree of airway constriction throughout the bronchial tree.

Very little is known about the effect of posture on $Z_{\text{cw}}$, likely due to the difficulties in measuring chest wall flow in the supine position. In the supine position, we found that both $Z_{\text{in}}$ and $Z_{\text{tr}}$ presented a substantial increase in their low-frequency components (Figs. 4 and 7). Interestingly, even if $Z_{\text{tr}}$ $R$ at the lowest frequency increased from 6.9 to 10.8 cmH$_2$O·l$^{-1}$·s$^{-1}$ passing to the supine position, the relative contribution of the $Z_{\text{cw}}$ to $Z_{\text{tr}}$ was almost the same as in the seated position at the lowest frequency (63%), whereas it slightly increased at the highest frequency (36%). In 1993, Barnas et al. (7) measured the effect of posture on $Z_{\text{cw}}$ by using a sinusoidal forcing at 0.2 Hz. They reported a baseline $Z_{\text{cw}}$ value and changes in $Z_{\text{cw}}$ due to posture that were smaller than ours. However, they used $V_{ao}$ to compute $Z_{\text{cw}}$ instead of chest wall flow, and, therefore, the effect of upper airway and cheek shunting might be responsible for these differences.

Compartmental analysis. The OEP technique allowed us to partition the volume changes of chest wall into compartments. Such partitioning allowed us to subdivide $Z_{\text{tr}}$ into its compartmental pathways. Implicit in the definitions of the compartmental impedances of Table 1 is that all of the pathways are independent. However, as Loring and Mead (25) first pointed out, these compartments are not mechanically independent, as $R_{\text{ca}}$ and $A_{\text{b}}$ are submitted to the same pressure (abdominal pressure) and $R_{\text{ca}}$ and $R_{\text{cp}}$ are mechanically linked by anatomic structures (the ribs and the sternum). Several models have been suggested that include the effect of the different pressures acting on the rib cage as well as the transmission of forces through mechanically coupled structures (12, 19, 21, 31, 43). In our study, the measurements occur with relaxed chest wall. Hence, we decided to ignore the coupling between compartments, as its effect on impedance estimation should be negligible.

Figure 5 shows that, in the seated position, the $R_{\text{cp}}$ and the $A_{\text{b}}$ displayed similar $R$ and $E$ values, making the whole chest wall behave like a single, homogeneous structure. The results in the supine position are completely different, with the $R_{\text{cp}}$ almost doubling both its $R$ and $E$ and the $A_{\text{b}}$ slightly reducing them. Because the two pathways are in parallel, the total $Z_{\text{tr}}$ in the supine position displays only a small increase in $R$ and essentially the same $E$ as in the seated position. Nevertheless, the chest wall behaves more inhomogeneously in the supine position, with the $A_{\text{b}}$ being the most important compartment in determining total respiratory system mechanics. Therefore, in the supine position, changes in the abdominal mechanical properties have a much bigger effect on the total respiratory mechanics than in the seated position. This may have important implications regarding patient care in the intensive care unit. Such patients may experience abdominal distension from various origins (39) (as, for example, surgical acute respiratory distress syndrome patients) (35), which increases abdominal

\begin{figure}[h]
\centering
\includegraphics[width=0.8\textwidth]{fig8.png}
\caption{Ratio between the resistance of the pulmonary rib cage ($R_{\text{rc,p}}$) and the abdominal pathway ($R_{\text{ab}}$) in the seated (circles) and the supine (triangles) position before (solid symbols) and after (open symbols) bronchoconstriction. Values $>1$ indicate that the pulmonary rib cage presents a larger resistance than the abdomen. The higher the value, the greater the $R_{\text{rc,p}}$ and, therefore, the greater the amount of in-phase flow displaced by the abdomen for a given $P_{ao}$.}
\end{figure}
stiffness (33). Based on our results, this condition will greatly amplify the Ztr and Zin in the supine position, which will make spontaneous breathing and mechanical ventilation far more arduous.

This increased influence of the Ab in the supine position is likely due to the direction of the gravitational forces relative to the physical support of the thorax. Specifically, in the supine position, most of the weight of the thorax is supported from the back of the subject, with the ribs supported directly by the bed and, therefore, less free to move and expand the rib cage. Conversely, the Ab has no rigid structure around the back (excluding the spinal chord), and the most important effect changing to the supine position is the change of direction of gravity on the abdominal contents. Thus, in supine, the abdominal weight is no longer stretching the abdominal walls, and it is now pushing on the back (where it is supported by the bed).

The result is that the abdominal contents are moved cranially, pushing the diaphragm up and lowering the lung volume (1). In this condition, the Ab has more freedom to move, as shown by its lower R and E.

In summary, under the hypothesis of a uniformity of pleural swings around the lung, the combination of OEP and OVW allows us to noninvasively track alterations in regional chest wall mechanics, providing potentially useful information for patients with restrictive disease, those in the intensive care unit, and for postoperative patients, especially those undergoing upper abdominal surgery.

Effect of bronchoconstriction. Bronchoconstriction had two major effects on the respiratory system. The most important one was on the Zl, where heterogeneous constriction of the airways has been shown to induce an increase of both mean level and frequency dependence of R. The second effect is on the chest wall configuration, with dynamic hyperinflation causing the end-expiratory Vcw to increase differently in each compartment and thus changing the range of the compartmental pressure-volume curve over which the subject is being ventilated.

As shown in Fig. 6, after MCh, lung R and E markedly increased with minor effects on Zcw. The absence of changes of Zcw indicates that, even if the subjects hyperinflated differently between the compartments when constricted (Table 2), this change was not sufficient to induce significant changes on the total mechanical properties of the chest wall. Figure 6 also shows that, because Zcw did not change significantly, the impact of bronchoconstriction on Zl is accurately reported by the changes of Ztr, as it is the series of the two respiratory structures. Thus the combination of OVW and OEP also provides a noninvasive tool for measuring the impact of bronchoconstriction on the lung without requiring the esophageal balloon. Conversely, the effect of bronchoconstriction is underestimated by Zin, as shown in Fig. 7, where the real part of Zin lies below Ztr, particularly in the supine position where the level of constriction is higher. The amplified sensitivity of Ztr to MCh compared with Zin in the supine position is consistent with a greater effect of airway wall shunting on Zin during the challenge.

It is interesting to compare how the compartmental pathway impedances are modified by MCh in the seated vs. the supine positions. In Fig. 8, the ratios between the real part of the RCP and the abdominal pathways are presented in the two postures before and after MCh. Values greater than one indicate that the RCP presents a larger R than the Ab. The higher the value, the greater the RCP R and, therefore, the greater the amount of in-phase flow displaced by the Ab for a given Pao. Bronchoconstriction induced different effects on the two postures. In the seated position, it markedly changed the frequency dependence of the compartmental contributions. Here, at high frequency, the RCP R was ~0.8 times the abdominal R, both before and after MCh, whereas, at low frequency, it was 1.1 times in control conditions and to 1.5 times after MCh. In the supine position, the contributions of the two compartments presented smaller frequency dependence than in the seated position, with the RCP R being 2.5 times the abdominal one at the lowest frequency and 1.7 at the highest frequency. After bronchoconstriction, even if the level of constriction reached in the supine position was higher than in the seated one, the relative contribution of the RCP was 2.5 at the lowest frequency and 1.3 at the highest, suggesting that posture affects the relative changes of compartmental mechanical properties differently. It is still unclear whether the differences observed are the results of different mechanical properties of the chest wall compartments or whether the heterogeneity of bronchoconstriction throughout the lung has a measurable effect on macroscopic regional mechanics, resulting in regional heterogeneity of pleural pressure swings.

Summary. We introduced a new, reliable method for measuring the mechanical impedances of the respiratory system and its components at low frequencies. This method exploits the OVW approach to provide ventilation and multifrequency forcing simultaneously while using the OEP method to measure chest wall flow, both globally and locally. Hence, we can estimate respiratory Ztr noninvasively. These data are appropriately sensitive to changes in posture, showing evidence of added R and decreased volume in supine vs. seated position. The Ztr are also highly sensitive to heterogeneous lung constriction, more so than Zin, as the former is minimally distorted by shunting of flow into alveolar gas and upper airway walls. Local impedances show that, during heterogeneous bronchoconstriction and at typical breathing frequencies, the contribution of the Ab becomes amplified relative to the rib cage. A similar redistribution occurs when passing from seated to supine. These data suggest that the OEP-OVW approach for measuring Ztr could noninvasively track important lung and respiratory clinical conditions, even in subjects who cannot perform specific maneuvers (e.g., neuromuscular disease, critically ill or postoperative patients, spinal chord injury patients) as the only requirement is the ability to relax the respiratory muscles. Thus applications can range from routine evaluation of airway hyperreactivity in asthmatic subjects to critical conditions in the supine position during mechanical ventilation.

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