RESPIRATORY INPUT IMPEDANCE DURING HIGH FREQUENCY OSCILLATORY VENTILATION

D. NAVAJAS AND R. FARRÉ

Pressure and flow measured at the airway opening are useful to determine ventilator settings during high frequency oscillatory ventilation (HFOV). Nevertheless, these data do not provide direct information on the changes in pressure throughout the ventilatory system. To monitor HFOV adequately, it would be of interest to know how pressure changes throughout the airways and, in particular, to estimate the amplitude of pressure oscillation in the alveolar region. This information is difficult to obtain in practice, as alveolar pressure ($P_A$) cannot be measured directly in patients and, moreover, pressure at the airway opening ($P_{ao}$) does not indicate directly the value of $P_A$. In this context, measurement of input impedance ($Z_{rs}$) may be useful for monitoring the ventilatory system [1]. First, $Z_{rs}$ determines mechanical load and can be used to predict the amplitude of pressure at the airway opening ($|P_{ao}|$) for different conditions of ventilation. Indeed, impedance amplitude ($|Z_{rs}|$) at each frequency represents the driving pressure amplitude ($P_{ao}$) necessary to deliver a flow amplitude ($|V_{ao}|$) of 1 litre s$^{-1}$ to the ventilatory system. Second, the analysis of $Z_{rs}$ data with an adequate model of the ventilatory system may be useful to predict the amplitude of alveolar pressure oscillations ($|P_A|$) from measurements at the airway opening. Unfortunately, a precise determination of $P_A$ cannot be carried out without difficulty, as the use of complicated mechanical models would be required to account for the complexity of ventilatory mechanics during HFOV [2-4].

In this study we have analysed how the information provided by $Z_{rs}$ may be used for monitoring the mechanics of the ventilatory system during HFOV. We measured $Z_{rs}$ in dogs at frequencies of 0.125–32 Hz to assess the mechanical load of the ventilatory system under linear conditions for oscillations which, for any given frequency, were of constant flow and constant volume. The $Z_{rs}$ data between 4 and 32 Hz were analysed using a linear resistance–inertance–elastance model to estimate $|P_A|$ and the $|P_A : P_{ao}|$ ratio. This model was applied also to $Z_{rs}$ measured after infusion of histamine. Finally, we analysed the effect on $|P_{ao}|$ of airways non-linearities at high flow amplitudes.

METHODS

Dogs

We studied six anaesthetized, paralysed mongrel dogs (11-20 kg). Anaesthesia was induced with pentobarbitone 30 mg kg$^{-1}$ i.v. and
paralysis with pancuronium $0.05-0.15$ mg kg$^{-1}$ i.v. After tracheotomy, a cannula (40 mm in length, i.d. between 13 and 17 mm, depending on the dog's size) was fitted into the trachea.

**High frequency oscillatory ventilation**

The electromechanical system used for HFOV is shown in figure 1. It was based on a servo-controlled linear motor with a coil (7 cm of maximum stroke) attached to rubber bellows of diameter approximately 10 cm. The linear motor was driven by a power amplifier (300 W). The system was servo-controlled by measuring the movement of the bellows by means of a displacement transducer (LVDT AC/100, Sangamo) coupled mechanically to the coil of the motor. A pneumatic low-pass filter consisting of a tube with a diameter of 2 cm and 150 cm in length was incorporated into the system ("bias tube" in figure 1). A bias flow of 0.2 litre s$^{-1}$ was applied at the distal end of the cannula.

To measure flow at the airway opening, the outlet of the bellows was attached to a mesh wire screen pneumotachograph (0.1 hPa s litre$^{-1}$) coupled to a differential pressure transducer ($\pm 2$ hPa, Validyne MP-45). This transducer had a common mode rejection ratio greater than 65 dB at 32 Hz. A similar transducer ($\pm 50$ hPa) was connected to the cannula to measure the tracheal pressure. Pressure and flow signals were low-pass filtered (analogue method) (8 poles, Butterworth, 3 dB at 32 Hz) and sampled at a rate of 128 Hz. The frequency response of the measuring system was determined using a reference impedance [5].

The system was controlled by means of a microcomputer (PS/30, IBM). It generated a sinusoidal HFOV signal to drive the power amplifier through a digital-to-analogue converter, and it acquired pressure and flow signals through an analogue-to-digital converter.

**Impedance measurements**

To measure $Z_{rs}$ we used the same electromechanical system as shown in figure 1. In this case, however, the signal generated by the computer to drive the linear motor was pseudorandom. This signal was generated by adding simultaneously 20 sinusoidal components with frequencies $0.125, 0.25, 0.5, 1, 2, 4, \ldots, 32$ Hz. The duration of this excitation signal and there-
Procedure

The dog's lungs were ventilated with HFOV at a frequency of 12 Hz with a tidal volume (VT) of 4 ml kg\(^{-1}\). To measure respiratory impedance under linear conditions, the 12-Hz signal to drive HFOV was replaced by the low amplitude pseudorandom signal (approximately 0.4 litre s\(^{-1}\) peak-to-peak). During impedance measurements, bias flow was interrupted and the bias tube occluded. HFOV was restored at the end of the impedance measurement. Four dogs were submitted to a histamine challenge by injection of a continuous infusion of histamine solution until the maximum airway pressure during conventional ventilation increased by 30–60%. Zrs measurements were performed after reaching a steady-state which was maintained for at least 5 min.

RESULTS AND DISCUSSION

Mechanical load of the ventilatory system under linear conditions

Respiratory impedance measurements were performed with a low amplitude excitation signal. Therefore, Zrs characterizes the mechanical response of the ventilatory system under almost linear conditions. The mean value of impedance amplitude for the six dogs is shown in figure 3. |Zrs| decreased from mean 86.4 (SD 33.9) hPa s litre\(^{-1}\) at 0.125 Hz to 7.2 (2.5) hPa s litre\(^{-1}\) at 2 Hz. Subsequently, |Zrs| decreased slightly, to a minimum of 2.2 (0.6) hPa s litre\(^{-1}\) at 10 Hz (frequency of resonance). At higher frequencies |Zrs| increased slightly up to 5.7 (1.0) hPa s litre\(^{-1}\) at 32 Hz. This frequency dependence pattern is in agreement with previous data [8, 9]. The marked negative frequency dependence of |Zrs| at the lowest frequencies may be attributed to tissues elastance. At the resonant frequency, the elastic load is compensated by the inertial load and thus |Zrs| is determined mainly by the resistive properties of the ventilatory system. Finally, above the resonant frequency the mechanical load is determined progressively by respiratory inertance.

The mechanical load of the ventilatory system at different frequencies for constant VT is characterized by the amplitude of the complex elastic modulus (|Trs|) [10]. |Trs| is the ratio of the simple model we were able to estimate ventilatory variables with a simple algorithm [7].

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The mechanical load of the ventilatory system at different frequencies for constant VT is characterized by the amplitude of the complex elastic modulus (|Trs|) [10]. |Trs| is the ratio of the
amplitude of pressure oscillation at the airway opening and the amplitude of volume oscillation ($V_T/2$). $|\Delta r_s|$ may be calculated as $|\Delta r_s| = 2\pi f |Z_{rs}|$ as, for sinusoidal oscillations, $|\Delta P_{ao}| = \pi f V_T$. Figure 3 shows the mean $|\Delta r_s|$ obtained from our $|Z_{rs}|$ data. These results indicate that the mechanical load of the ventilatory system during ventilation at constant $V_T$ is similar from very low frequencies (0.125 Hz) to the frequency of resonance. At higher frequencies the load increased sharply.

### Table I. Parameters of the resistance-inertance-elastance (R–I–E) model

<table>
<thead>
<tr>
<th>Dog No.</th>
<th>$R$ (hPa s litre$^{-1}$)</th>
<th>$I$ (Pa s$^2$ litre$^{-1}$)</th>
<th>$E$ (hPa litre$^{-1}$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>2.55</td>
<td>2.50</td>
<td>116.4</td>
</tr>
<tr>
<td>2</td>
<td>3.39</td>
<td>2.23</td>
<td>192.3</td>
</tr>
<tr>
<td>3</td>
<td>2.61</td>
<td>2.00</td>
<td>124.1</td>
</tr>
<tr>
<td>4</td>
<td>2.95</td>
<td>2.26</td>
<td>123.3</td>
</tr>
<tr>
<td>5</td>
<td>2.91</td>
<td>3.17</td>
<td>61.7</td>
</tr>
<tr>
<td>6</td>
<td>2.09</td>
<td>2.10</td>
<td>109.9</td>
</tr>
<tr>
<td>Mean</td>
<td>2.75</td>
<td>2.38</td>
<td>121.3</td>
</tr>
<tr>
<td>SD</td>
<td>0.44</td>
<td>0.42</td>
<td>41.9</td>
</tr>
</tbody>
</table>

To estimate the pressure change across the airways, we assumed that $R$ and $I$ of the model in figure 2 represent the resistance and inertance of the airways. Accordingly, airways impedance is $R + j2\pi f I$ and therefore alveolar pressure amplitude may be estimated as:

$$|P_A| = |Z_{rs} - (R + j2\pi f I)| \cdot \pi f V_T$$

We computed $|P_A|$ from this equation by taking the mean measured impedance (the amplitude is shown in figure 3) and mean $R$ and $I$ variables (table I) for a constant $V_T = 50$ ml. The results are shown in figure 4 (solid lines). The $|P_A|$ estimated was almost constant with frequency as, in the model used, $|P_A|$ is determined mainly by the elastance parameter. This frequency dependence of $|P_A|$ is consistent with alveolar pressures reported in excised dog lungs [2]. This suggests that the linear series R–I–E model could lead to reasonable estimation of $|P_A|$. Moreover, parameters in this model can be computed almost in real time by using a linear algorithm with single solution, which is useful for monitoring. Figure 4 (solid lines) also includes the ratio $|P_A|/P_{ao}$. This was close to 1 at frequencies up to resonant, and decreased markedly down to 0.25 at 20 Hz. Figure 4 suggests, therefore, that at frequencies up to approximately resonance, $|P_A|$ is similar to $|P_{ao}|$, but becomes progressively smaller as frequency increases.

When $Z_{rs}$ of the four dogs submitted to the histamine challenge were analysed with the R–I–E model, we obtained $R = 6.07$ (1.56) hPa s litre$^{-1}$, $I = 2.60$ (1.24) Pa s$^2$ litre$^{-1}$ and $E = 259.2$ (114.0) hPa litre$^{-1}$. The large increases in $R$ and $E$ found are consistent with published data after histamine challenge in dogs [11-14]. The values of $|P_{ao}|$, $|P_A|$ and $|P_A|/P_{ao}$ which we computed after histamine infusion for the same $V_T$ as in the basal conditions (50 ml) are included in dotted lines in
Fig. 4. Estimated amplitudes of pressure oscillation for a constant tidal volume of VT = 50 ml in basal conditions (---) and after infusion of histamine (----). |Pao| = Amplitude of pressure at the airway opening; |PA| = alveolar pressure amplitude.

|Pao| was greater after histamine because of the increase in load impedance (R, I and E increased). As in the basal conditions, the estimated |PA| was almost independent of frequency. The highest values of |PA| after histamine could be attributed to the increase in elastance. Figure 4 shows that |PA:Pao| was smaller after histamine below the resonant frequency. Nevertheless, at higher frequencies |PA:Pao| was similar to that at basal conditions, because at high frequencies the mechanical load is determined progressively by inertance, which was the variable that exhibited the smallest change after infusion of histamine.

Fig. 5. Estimated amplitudes of pressure oscillation for a constant VT = 50 ml, when assuming linear conditions (---) or when assuming a non-linear component of airways resistance (----) (K₂ = 0.5 hPa litre⁻¹ s⁻¹).

**Modelling airways non-linearities**

The data discussed above were derived from Zrs values obtained by exciting the ventilatory system with a low amplitude signal, that is under almost linear conditions. In contrast, during HFOV, flow amplitudes are greater. As a consequence, the pressure change across the airways may exhibit a considerable non-linear component which increases ventilatory mechanical load. To analyse this non-linear behaviour, we modelled the effective airways resistance (R*) as in the Rohrer equation (R* = R + K₂. V). Accordingly, when submitted to a sinusoidal flow, the effective impedance (Zrs*) of the respiratory system is [15]:

\[ Zrs* = Zrs + (8/3\pi). K_2. \dot{V}_0 \]  

where \( \dot{V}_0 \) is the amplitude of the oscillatory flow and \( K_2 \) is the non-linear coefficient of airways resistance. This equation shows that \( K_2 \) may be determined from impedance measured under linear conditions, Zrs, and from the effective impedance, Zrs*, measured with a sinusoidal oscillation of high flow amplitude. For instance, in dog No. 6 the real part of measured Zrs at 12 Hz was 2.35 hPa s litre⁻¹, and the real part of effective impedance Zrs* at 12 Hz with a flow
amplitude of 1.95 litre s\(^{-1}\) was 3.20 hPa litre\(^{-1}\). Therefore, \(K_2 = 0.51\) hPa litre\(^{-2}\) s\(^{2}\) was calculated from equation (3).

Knowledge of \(K_2\) permits us to estimate mechanical load at high flow amplitudes and, consequently, to predict the amplitude of pressure oscillation at the airway opening. Figure 5 shows (dashed lines) the \(|P_{ao}|\) estimated from Zrs of figure 3 for \(V_T = 50\) ml, assuming linear conditions \((K_2 = 0)\) and in solid lines the value estimated assuming \(K_2 = 0.5\) hPa litre\(^{-2}\) s\(^{2}\). Below 8 Hz, the flow amplitude corresponding to \(V_T = 50\) ml is less than 1.3 litre s\(^{-1}\) and thus the difference between \(|P_{ao}|\) computed with \(K_2 = 0.5\) hPa litre\(^{-2}\) s\(^{2}\) and \(P_{ao}\) computed under linear conditions is less than 1 hPa. In contrast, at 32 Hz the flow amplitude for the same \(V_T\) is 5.0 litre s\(^{-1}\) and thus the difference in \(|P_{ao}|\) is 8.0 hPa (fig. 5). Figure 5 also shows the change in \(|P_{ao}:P_{ao}|\) from linear (dashed line) to non-linear (solid line) conditions. When the airways non-linearity was taken into account, \(|P_{ao}:P_{ao}|\) was smaller than under linear conditions over the entire frequency range (4–32 Hz).

In conclusion, measuring Zrs and interpreting it in terms of a series model including a non-linear term for resistance is useful to characterize the mechanical load of the ventilatory system during HFOV and to estimate the amplitude of alveolar pressure oscillation. Moreover, Zrs computation and the parameter estimation of this model can be carried out easily and quickly during HFOV.

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