Effect of the Ti/Te ratio on mean intratracheal pressure in high-frequency oscillatory ventilation

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Thome, Ulrich, and Frank Pohlandt. Effect of the Ti/Te ratio on mean intratracheal pressure in high-frequency oscillatory ventilation. J. Appl. Physiol. 84(5): 1520–1527, 1998.—In high-frequency oscillatory ventilation (HFOV), an adequate mean airway pressure is crucial for successful ventilation and optimal gas exchange, but air trapping cannot be detected by the usual measurement at the y piece. Intratracheal pressures produced by the high-frequency oscillators HFV-Infantstar (IS), Babylong 8000 (BL), and the SensorMedics 3100A (SM) [the latter with either 30% (SM30) or 50% (SM50) inspiratory time] were investigated in four anesthetized tracheotomized female piglets that were 1 day old and weighed 1.6–1.9 kg (mean 1.76 kg). The endotracheal tube was repeatedly clamped while the piglets were ventilated with an oscillation frequency of 10 Hz, and the airway pressure distal of the clamp was recorded as a measure of average intrapulmonary pressure during oscillation. Clamping resulted in a significant decrease of mean airway pressure when the piglets were ventilated with SM30 (0.86 cmH2O), BL (0.66 cmH2O), and IS (0.71 cmH2O), but airway pressure increased by a mean of 0.76 cmH2O with SM50. Intratracheal pressure, when measured by a catheter pressure transducer at various oscillation frequencies, was lower than at the y piece by 0.4–0.9 cmH2O (SM30), 0.3–3 cmH2O (BL), and 1–4 cmH2O (IS) but was 0.4–0.7 cmH2O higher with SM50. We conclude that the inspiratory-to-expiratory time (Ti/Te) ratio influences the intratracheal and intrapulmonary pressures in HFOV and may sustain a mean pressure gradient between the y piece and the trachea. A Ti/Te ratio <1:1 may be useful to avoid air trapping when HFOV is used.

SUFFICIENTLY HIGH MEAN AIRWAY PRESSURE (MAP) in high-frequency oscillatory ventilation (HFOV) has been shown to be crucial for adequate recruitment of the infant's lung and for the achievement of optimum results: pulmonary function and compliance were improved, lung damage was reduced, and the effectiveness of exogenous surfactant was prolonged when lung volume was optimized in surfactant-deficient rabbits (14, 20) and premature baboons (12). Similarly, HFOV improved the outcome and reduced the complications only in those clinical studies in which the ventilatory strategy aimed at an optimization of lung volume (10, 17) but not in those with a strategy involving lower mean pressures (6, 16). Intratracheal pressure and lung volume were important determinants of oxygenation in the HFOV of dogs (25), rabbits (26), and preterm infants (9) with respiratory failure. Currently available high-frequency ventilators measure MAP at the y piece of the ventilator circuit, which may not reflect true intrapulmonary pressure. In animal experiments, intratracheal mean pressure has been found to be lower than at the y piece in HFOV with the SensorMedics 3100A oscillator (SM) set to an inspiratory-to-expiratory time (Ti/Te) ratio of 1:2.3 (15) or 1:2 (29). When SM and other devices were used with a Ti/Te ratio of 1:1, however, lung hyperinflation was demonstrated (1, 5, 8, 15, 18, 22, 24). Significant gas trapping occurred in high-frequency jet ventilation, which lacked an active exhalation, but not in HFOV (2).

Our study was undertaken with three commercially available neonatal high-frequency oscillators to investigate whether gas trapping occurs and whether its occurrence depends on the different techniques used to generate the positive- and negative-pressure swings. The oscillation of two of the devices was always asymmetrical, i.e., the inspiratory phase was shorter than the expiratory phase. In one of the devices, an adjustable Ti/Te ratio enabled us to investigate the influence of different Ti/Te ratio settings (1:2 or less, asymmetrical, and 1:1, symmetrical) on gas trapping. Finally, intratracheal waveform recordings were obtained to estimate how low-pass filtering affects the pressure waveforms in passing through the endotracheal tube. A potential distortion of our measurements due to the Bernoulli effect was excluded.

METHODS

Subjects. Female domestic piglets that were 1 day old and weighed 1.6–1.9 kg (mean 1.76 kg) were sedated by intramuscular injection of 8 mg atazepam, anesthetized by intravenous injection of 10 mg medetomidine, and endotracheally intubated (tube diameter 3.5 mm). Anesthesia was maintained by continuous infusion of pentobarbital sodium as needed to suppress spontaneous movements and respiratory efforts, together with 5 ml/h electrolyte solution with 10% glucose. Body temperature was kept between 36 and 37°C by placing a heating pad under the animal and a radiator above it. The experiments were performed according to institutional guidelines and approved by the local committee for animal research.

Instrumentation. The trachea was exposed and ligated proximal to the end of the endotracheal tube to exclude leaks. A cardiac catheter (5-Fr) with a piezoresistive pressure transducer on its tip (model PC 350, Millar Instruments, Houston, TX) was inserted into the trachea through a small hole near the end of the endotracheal tube—thereby not reducing the endotracheal tube's lumen—and advanced ~2 cm. The estimated position of the tip was ~1 cm proximal of the carina. The insertion hole was sealed around the catheter with cyanoacrylate tissue glue. A Statham pressure transducer and a water manometer (accuracy 0.1 cmH2O, compliance 0.09 ml/cmH2O) were connected between the y piece of the ventilator tubing system and the endotracheal tube (see Fig. 1). The airway diameter at the connection site was 5 mm. The intratracheal and extratracheal pressure transducers were calibrated by opening the endotracheal tube to admit
room air (zero pressure) and applying constant pressure steps with the oscillation switched off (continuous positive airway pressure), by using the water manometer as a calibration standard. The animals were apneic throughout. The right common carotid artery was cannulated for monitoring blood pressure and obtaining samples for blood-gas analyses. Oxygenation was monitored with a pulse oximeter (Nellcor) attached to one lower leg, and inspired oxygen fraction was adjusted to maintain oxygen saturation >95%.

Measurements. Measurements were performed with HFOV only, without additional conventional breaths, by using a SensorMedics 3100A (SensorMedics, Yorba Linda, CA) set to either 50% (SM50) or to 30% inspiratory time (SM30) (Ti/TE ratio 1:1 or 1:2.3, respectively); a Babylog 8000 (BL; HFOV software version 4.0, Drägerwerk, Lübeck, Germany); and an HFV-Infantstar (IS; software version 4.0, Nellcor Puritan Bennett, Carlsbad, CA). The Ti/TE ratio of the BL was 1:2 and was not changeable by the user. The IS had a fixed inspiratory time of 18 ms in the high-frequency mode; the expiratory time and the Ti/TE ratio were thus dependent on inspiratory time of 18 ms in the high-frequency mode; the Ti/TE ratio was determined by measuring the duration of inspiratory and expiratory flows and calculating the quotient. At the end, the animal was euthanized with a pentobarbital sodium overdose. All measurements were performed with HFOV only, without additional conventional breaths, by using a SensorMedics 3100A (SensorMedics, Yorba Linda, CA) set to either 50% (SM50) or to 30% inspiratory time (SM30) (Ti/TE ratio 1:1 or 1:2.3, respectively); a Babylog 8000 (BL; HFOV software version 4.0, Drägerwerk, Lübeck, Germany); and an HFV-Infantstar (IS; software version 4.8, Nellcor Puritan Bennett, Carlsbad, CA). The Ti/TE ratio of the BL was 1:2 and was not changeable by the user. The IS had a fixed inspiratory time of 18 ms in the high-frequency mode; the expiratory time and the Ti/TE ratio were thus dependent on the oscillation frequency (f).

Before the start of the measurements, the MAP that produced the best possible oxygenation with the lowest possible inspired oxygen fraction without compromising arterial blood pressure in each animal was determined by using SM30. This MAP was subsequently used throughout the experiment, and its values ranged from 6 to 12 cmH2O in all animals. To use equally effective oscillation amplitudes in all ventilators, the oscillation amplitude was adjusted for each ventilator and animal to result in arterial PCO2 (PaCO2) values between 35 and 45 Torr at 5 l/min (23) was inserted between the y piece and the endotracheal tube (see Fig. 1) for simultaneous registration of flow and pressure waveforms of each oscillator. The Ti/TE ratio was determined by measuring the duration of inspiratory and expiratory flows and calculating the quotient. At the end, the animal was euthanized with a pentobarbital sodium overdose. All measurements were performed with HFOV only, without additional conventional breaths, by using a SensorMedics 3100A (SensorMedics, Yorba Linda, CA) set to either 50% (SM50) or to 30% inspiratory time (SM30) (Ti/TE ratio 1:1 or 1:2.3, respectively); a Babylog 8000 (BL; HFOV software version 4.0, Drägerwerk, Lübeck, Germany); and an HFV-Infantstar (IS; software version 4.0, Nellcor Puritan Bennett, Carlsbad, CA). The Ti/TE ratio of the BL was 1:2 and was not changeable by the user. The IS had a fixed inspiratory time of 18 ms in the high-frequency mode; the expiratory time and the Ti/TE ratio were thus dependent on the oscillation frequency (f).

To study intrapulmonary pressure in relation to MAP at the y piece, the endotracheal tube was cross-clamped for 3 s proximal to the pressure sensors (Fig. 1), and the pressure change distal of the clamp, a measure of the pressure at the y piece, the endotracheal tube was cross-clamped for 3 s proximal to the pressure sensors (Fig. 1), and the pressure change distal of the clamp, a measure of the pressure difference between the y piece and the lung during oscillation, was read from the water manometer and simultaneously registered by the chart recorder. Leaks were excluded by observing a pressure plateau on the chart recorder after the clamp was closed. The animals were apneic throughout the experiments. Single clamping resulted in either a pressure rise or a pressure drop at the y piece, depending on the instant in the oscillatory cycle at which the clamping occurred (5). To obtain a mean pressure change, clamping was repeated 20 times per animal with each of SM50, SM30, BL, and IS in random order, and mean pressure changes for each animal and ventilator setting were calculated. From these individual means, mean values and SDs across all four animals were obtained and analyzed with Student’s t-test for unpaired observations.

Thereafter, the mean pressure difference between y piece and trachea was determined by using the proximal and intratracheal sensors. The oscillators were interchanged in random order, and the preset MAP and oscillation amplitudes were not changed from the first experiment. With each oscillator, measurements were carried out at the following f values: 1, 3, 5, 7, 10, 12, 15, 17, and 20 Hz, as far as supported by the respective oscillator, always starting with the lowest frequency and increasing stepwise. To test whether the pressure in the trachea differed significantly from that at the y piece when the animals were ventilated with SM50, SM30, BL, and IS, the areas (integrals) between the curves of intratracheal pressure vs. f and the pressure at the y piece vs. f in the frequency range between 5 and 15 Hz were calculated for each animal by using the trapezoid rule. Median values were obtained for SM50, SM30, BL, and IS and were tested with the Wilcoxon signed-rank test for deviation from zero (P < 0.05).

Finally, a pneumotachograph, linear up to 15 l/min, with a resistance of 1.1 kPa·s·l−1 at 5 l/min (23) was inserted between the y piece and the endotracheal tube (see Fig. 1) for simultaneous registration of flow and pressure waveforms of each oscillator. The Ti/TE ratio was determined by measuring the duration of inspiratory and expiratory flows and calculating the quotient. At the end, the animal was euthanized with a pentobarbital sodium overdose. All measurements were recorded with a multigraph (Hellige, Freiburg, Germany).

Calculation of the Bernoulli effect. The measurement of the mean pressure at the y piece by a lateral-pressure tap may be distorted by the Bernoulli effect, because only the static component (Pst) of the total pressure (Pt) is being measured. According to the Bernoulli formula, Pt underestimates Pst by 1/2ρv2 (3).

\[ P_t = P_{st} + \frac{1}{2} \rho v^2 \]  
\[ \text{(1)} \]

where \( \rho \) is gas density and \( v \) is average flow velocity across the tube. To calculate the dynamic pressure component (Pdyn)

\[ P_{dyn} = P_t - P_{st} = \frac{1}{2} \rho v^2 \]  
\[ \text{(2)} \]
the flow velocity \( v \) can be estimated with sufficient accuracy by dividing the average flow \( \bar{V} \) by the cross-sectional area \( A \), as shown by Simon et al. (24)

\[
v = \frac{\bar{V}}{A} = \frac{\bar{V}}{\pi r^2} = \frac{\bar{V}}{\pi d^2} = \frac{4\bar{V}}{\pi d} \tag{3}
\]

where \( r \) is radius and \( d \) is diameter. If Eq. 3 is inserted into Eq. 2, we get

\[
P_{\text{dyn}} = \frac{8}{\pi^2} \frac{\bar{V}^2}{d^4} \tag{4}
\]

For calculation of the \( P_{\text{dyn}} \), the average flow \( \bar{V} \) of each setting and ventilator was determined from the recorded flow waveforms. The integrals of inspiratory and expiratory flows were obtained, the absolute values of both added, and the sum was divided by the cycle length. The \( P \) was assumed to be 1.1 kg/m\(^3\) (36°C, altitude 500 m above sea level).

**RESULTS**

\( P_{\text{ACO}_2} \) values between 35 and 45 Torr were easily achieved only with SM\(_{50} \) and SM\(_{30} \). Despite maximum amplitude settings, \( P_{\text{ACO}_2} \) was >45 Torr in two of four animals with IS (range 35–51 Torr) and in all animals with BL (range 48–68 Torr). Clamping of the endotracheal tube resulted in a small but significant rise in mean pressure distal of the clamp with SM\(_{50} \), but with SM\(_{30} \), BL, and IS we observed a significant drop (\( P < 0.05 \); Fig. 2 and Table 1).

Intrathoracic pressure measurements at various \( f \) values, as shown in Fig. 3 and Table 2, yielded 0.4–0.7 cmH\(_2\)O higher mean pressures than at the y piece with SM\(_{50} \) (\( P < 0.05 \)). In contrast, with SM\(_{30} \), BL, and IS, the intrathoracic mean pressure was always lower than at the y piece by 0.4–0.9 cmH\(_2\)O (SM\(_{30} \), \( P < 0.05 \)). 0.3–3 cmH\(_2\)O (BL, \( P < 0.05 \)), and 1–4.7 cmH\(_2\)O (IS, \( P < 0.05 \)). Furthermore, the pressure difference between the tra- chea and the y piece was significantly greater with IS than with SM\(_{30} \) and BL (\( P < 0.05 \); Table 2).

The effective TI/TE ratio at the endotracheal tube connector, as determined from the durations of inspiratory and expiratory flow, was almost constant throughout the frequency range and close to the expected values with SM\(_{50} \) (expected 1:1.0, measured 1.0:1–1:1.1) and SM\(_{30} \) (expected 1:2.3, measured 1:2.1–1:2.4). The TI/TE ratio of IS was always greater than expected from the 18-ms inspiratory pulse duration and increased with frequency (1:1.25–1:1.3). The TI/TE ratio of BL also increased with frequency (1:2.2–1:1.3, Table 3).

Average flow rates during oscillation were between 23.5 and 84.5 ml/s. The Bernoulli effect (dynamic pressure) was <0.11 cmH\(_2\)O with all ventilators and settings (Table 3), and thus it did not significantly influence our results.

Flow and pressure waveforms at the y piece and intrathoracic pressure waveforms of SM\(_{50} \), SM\(_{30} \), BL, and IS at 10 Hz are shown in Fig. 4. The shapes of the inspiratory and expiratory waveforms of SM\(_{30} \) closely resembled each other, and so the oscillation was symmetrical. In SM\(_{30} \), BL, and IS, the inspiratory peak was shorter and higher than the expiratory trough. The shape of each inspiratory pulse was distinctly different from the shape of each expiratory pulse; the oscillation was thus asymmetric. All waveforms were strongly dampened in passing through the endotracheal tube, and most device-specific patterns, such as the incisurae of the SM waveforms or the short high peak of the IS waveform, were lost, leading to very similar intrathoracic pressure waveforms with all devices.

**DISCUSSION**

Two different experiments were performed in this animal study to measure mean intrapulmonary pressure and intrathoracic pressure during HFOV while comparing three high-frequency oscillators, one of them with two different TI/TE ratios. First, the average intrathoracic pressure change after repeated occlusion of the endotracheal tube enabled us to make an accurate estimate of the average intrapulmonary pressure

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Table 1. Mean pressure change in airway after cross-clamping the oscillation tube at 10-Hz oscillation frequency

<table>
<thead>
<tr>
<th>Device</th>
<th>Mean (n = 4)</th>
<th>95% Confidence Interval</th>
</tr>
</thead>
<tbody>
<tr>
<td>SensorMedics 3100A, TI/TE = 1:1</td>
<td>+0.76*</td>
<td>+0.14–1.38</td>
</tr>
<tr>
<td>SensorMedics 3100A, TI/TE = 1:2.3</td>
<td>-0.86*</td>
<td>-1.52–0.21</td>
</tr>
<tr>
<td>Babylog 8000</td>
<td>-0.66*</td>
<td>-0.89–0.43</td>
</tr>
<tr>
<td>HFV-Infantstar</td>
<td>-0.71*</td>
<td>-1.41–0.01</td>
</tr>
</tbody>
</table>

Clamping was repeated 20 times with each animal and ventilator setting, and mean pressure change (cmH\(_2\)O) was calculated separately for each animal. From these individual means, global mean values and confidence intervals listed above were obtained. TI/TE, inspiratory-to-expiratory time ratio. *Significantly different from zero (\( P < 0.05 \)).
during oscillation. This experiment did not require intratracheal sensors, and the pressure changes were measurable with a water manometer, which did not have to be calibrated. Depending on the phase in the oscillatory cycle in which the clamp was closed, pressure after clamping varied. To obtain representative mean values, clamping was repeated 20 times per animal and ventilator setting.

Second, intratracheal pressures over the complete frequency range provided by the devices were investigated with a catheter pressure transducer placed intratracheally. The direct measurement inside the trachea combined with the exceptional frequency response of the sensor (11) did not require correction of the signal as previously shown (4). The velocity-dependent Bernoulli effect may decrease the reading of the extratracheal pressure transducer during oscillation but not during clamping. Thus a pressure rise during clamping may be overestimated, and a pressure drop underestimated. In our setup, however, the Bernoulli effect should not have significantly altered the pressure measurements. Calculations of the Pdyn revealed values <0.11 cmH2O, which were much smaller than in experiments performed on adult dogs (24), probably because our piglets required much smaller tidal volumes and flow rates.

Different techniques were used for the generation of the oscillation in the high-frequency oscillators investigated. In the SM, a membrane electrically driven by a linear motor generated the inspiratory and expiratory phase of the oscillation. Both phases were thus active. Square waves were used at all times. The f range was 5–15 Hz, the TI/TE ratio was variable between 1:2.3 (SM30) and 1:1 (SM50). At the y piece, the effective TI/TE

Table 2. Areas (integrals) between the pressure vs. frequency curves in trachea and at the y piece in the oscillation frequency range of 5–15 Hz

<table>
<thead>
<tr>
<th>Subject No.</th>
<th>SM50</th>
<th>SM30</th>
<th>BL</th>
<th>IS</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>5.03</td>
<td>-10.64</td>
<td>-1.70</td>
<td>-13.81</td>
</tr>
<tr>
<td>2</td>
<td>+3.45</td>
<td>-7.51</td>
<td>-6.22</td>
<td>-13.51</td>
</tr>
<tr>
<td>3</td>
<td>+5.49</td>
<td>-15.81</td>
<td>-14.31</td>
<td>-21.57</td>
</tr>
<tr>
<td>4</td>
<td>+5.78</td>
<td>-8.02</td>
<td>-9.99</td>
<td>-15.45</td>
</tr>
<tr>
<td>Median</td>
<td>+5.26*</td>
<td>-9.33*</td>
<td>-8.11*</td>
<td>-14.63*</td>
</tr>
</tbody>
</table>

SM50, SensorMedics 3100A with TI/TE = 1:1, at inspiratory time (TI) = 50%; SM30, SensorMedics 3100A with TI/TE = 1:2.3, at TI = 30%; BL, Babylog 8000; IS, HFV-Infantstar. *Significantly different from zero (P < 0.05); †significantly different from the respective values of SM30 and BL (P < 0.05, multiple testing not taken into consideration).

Table 3. Measured TI/TE ratios, and the dynamic component of pressure at the y piece, as calculated from flow waveforms according to Eq. 4

<table>
<thead>
<tr>
<th>f, Hz</th>
<th>TI/TE</th>
<th>V, ml/s</th>
<th>Pdyn, cmH2O</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>SM50</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>1:1.1</td>
<td>60.4</td>
<td>0.052</td>
</tr>
<tr>
<td>5</td>
<td>1:1.1</td>
<td>76.1</td>
<td>0.083</td>
</tr>
<tr>
<td>7</td>
<td>1:1.1</td>
<td>68.3</td>
<td>0.067</td>
</tr>
<tr>
<td>10</td>
<td>1:1.0</td>
<td>81.3</td>
<td>0.094</td>
</tr>
<tr>
<td>12</td>
<td>1:1.0</td>
<td>74.9</td>
<td>0.08</td>
</tr>
<tr>
<td>15</td>
<td>1:1.1</td>
<td>67.5</td>
<td>0.065</td>
</tr>
<tr>
<td></td>
<td>SM30</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>1:2.4</td>
<td>46.8</td>
<td>0.031</td>
</tr>
<tr>
<td>5</td>
<td>1:2.3</td>
<td>59.0</td>
<td>0.05</td>
</tr>
<tr>
<td>7</td>
<td>1:2.1</td>
<td>60.8</td>
<td>0.053</td>
</tr>
<tr>
<td>10</td>
<td>1:2.1</td>
<td>57.8</td>
<td>0.048</td>
</tr>
<tr>
<td>12</td>
<td>1:2.3</td>
<td>59.5</td>
<td>0.081</td>
</tr>
<tr>
<td>15</td>
<td>1:2.3</td>
<td>58.2</td>
<td>0.048</td>
</tr>
<tr>
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<td>BL</td>
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<tr>
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<td>62.2</td>
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</tr>
<tr>
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<td>70.4</td>
<td>0.071</td>
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<tr>
<td>10</td>
<td>1:1.8</td>
<td>63.2</td>
<td>0.057</td>
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<tr>
<td>12</td>
<td>1:1.4</td>
<td>60.8</td>
<td>0.053</td>
</tr>
<tr>
<td>15</td>
<td>1:1.3</td>
<td>49.3</td>
<td>0.035</td>
</tr>
<tr>
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<td>1:1.3</td>
<td>49.6</td>
<td>0.035</td>
</tr>
<tr>
<td>20</td>
<td>1:1.3</td>
<td>49.8</td>
<td>0.035</td>
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<td>0.008</td>
</tr>
<tr>
<td>3</td>
<td>1:8.2</td>
<td>30.9</td>
<td>0.014</td>
</tr>
<tr>
<td>5</td>
<td>1:3.7</td>
<td>53.3</td>
<td>0.04</td>
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<td>74.3</td>
<td>0.079</td>
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<tr>
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<td>1:1.6</td>
<td>84.3</td>
<td>0.101</td>
</tr>
<tr>
<td>17</td>
<td>1:1.5</td>
<td>84.5</td>
<td>0.102</td>
</tr>
<tr>
<td>20</td>
<td>1:1.3</td>
<td>76.5</td>
<td>0.084</td>
</tr>
</tbody>
</table>

Pdyn, dynamic pressure component = 1/2 \( \rho v^2 \), where \( \rho \) is gas density = 1.1 kg/m³, and d is diameter = 5 mm (see Eq. 4). f, Oscillation frequency; V, average flow.
ratio was always close to the theoretical value and independent of f (Table 3). This device was by far the most powerful one in our study and achieved the desired PaCO2 values easily. Designed as a pure oscillator, it lacks the option of conventional ventilation.

The BL with HFOV software version 4.0 used an electrically driven membrane in the expiration valve to generate oscillations and a Venturi system similar to the IS for an active expiratory phase, although the effect of the latter may be less obvious in the waveform recordings (see Fig. 4C). The f range was 5–20 Hz, the Ti/Te ratio, which could not be modified by the operator, was dependent on f and between 1:2.2 (at 5 Hz) and 1:1.3 (at 20 Hz, Table 3). The maximum-pressure
amplitude was insufficient to produce the desired PaCO₂ values in our animals, but the ventilator performed well in our preterm infants who weighed less (data not shown).

In the IS, the opening of a high-pressure valve for short periods generated the inspiratory pulses. After each pulse, a Venturi system sucked air out of the ventilator circuit, resulting in an active expiratory phase. The f range was 1–22 Hz. As the pulse-on time was fixed to 18 ms, the TI/TE ratio was dependent on f and could not be adjusted independently. At 10 Hz, for instance, the theoretical TI/TE ratio was 1:4.6, at 20 Hz it was 1:1.8. Because of dampening, the inspiratory pulse became broader on its way through the heater and the tubing system, resulting in TI/TE ratios between 1:12.5 (at 1 Hz), 1:2.3 (at 10 Hz), and 1:1.3 (at 20 Hz) at the y piece. Thus the effective TI/TE ratio was only slightly smaller than that of SM₃₀ and BL at most frequencies (see Table 3). The new software version 83 with enhanced oscillation amplitude, which is now being built into the IS device, still uses a fixed pulse-on time of 18 ms; therefore, the results we obtained with software version 48 most likely apply to the software version 83 as well. The newer software may provide more efficient CO₂ removal.

Despite these technical differences, intratracheal pressure waveforms were similar. The square waves of SM and the sharp and short pressure peaks of IS, which contained more noisy overtones in their frequency spectrum, were subject to stronger low-pass filtering (dampening) in passing through the endotracheal tube than the smooth oscillations of BL (Fig. 4). The resulting intratracheal pressure waveforms, stripped of the overtones, lacked most device-specific patterns and looked very similar (see below). At 10 Hz, inspiration and expiration were incomplete in all devices, as shown by the constantly remaining pressure gradient between the y piece and the trachea at the end of both phases and by the steep angles at the intersections of the flow curve and the zero line. With SM₃₀, BL, and IS, the inspiratory time was shorter than the expiratory time, requiring a higher inspiratory peak than expiratory trough to compensate. This can be easily recognized if one draws a horizontal line, representing the mean pressure, through the extratracheal pressure curves in Fig. 4. B–D, so that the area between the line and the pressure curve is equal above and below it. The resulting inspiratory and expiratory waveforms were distinctly different; the oscillation was thus asymmetrical. Under these conditions, a significant pressure drop in the airways was observed after clamping the endotracheal tube, indicating that the lungs contained less air during oscillation than they would have had under continuous positive airway pressure of the same mean value at the y piece. This view is supported further by the direct measurements of the mean intratracheal pressure, which yielded significantly lower values than at the y piece. The difference was significantly greater with IS than with SM₃₀ or BL, probably because the IS had a smaller TI/TE ratio than SM₃₀ and BL at most frequencies (Table 3). The peculiar course of the curve of IS in Fig. 3 with a local minimum at 5 Hz and the minima of BL and IS at 20 Hz are difficult to explain. We speculate that the local minimum at 5 Hz may be caused by the extreme TI/TE ratio (1:3.7) of the IS at 5 Hz. Furthermore, the efficiency of the Venturi system generating the active expiration of BL and IS may be dependent on frequency.

BL had to be used with a slightly less effective oscillation-amplitude setting than the other devices, because higher amplitudes were not available. We speculate that the pressure difference between the Y piece and the trachea might have been higher if the ventilator had had higher oscillation amplitudes available.

Intratracheal and alveolar pressures lower than at the y piece were also found in adult rabbits (15) and juvenile pigs (weight 10–16 kg) (29). Our findings of a similar pressure gradient with two technically totally different oscillators (SM₃₀ and BL), which had only the Ti/Te ratio (1:2) in common, an even higher pressure gradient with the third device (IS), which had a more extreme Ti/Te ratio, plus the reports in the literature cited above lead to the hypothesis that it is the asymmetry of the oscillation with a shorter inspiration than expiration that is responsible for the lower intratracheal pressure, and not some other ventilator-specific effects.

The lower intratracheal pressure may be explained by the interaction of the strongly low-pass-filtering properties of the endotracheal tube (13) with the asymmetry of the oscillation. In analogy to electronics, the endotracheal tube and the lung can be viewed as a resistor and a capacitor, respectively, connected in series. This circuit is known as a low-pass filter. The endotracheal tube (resistor) limits flow (current), and thus delays filling and pressure (voltage) rise inside the lung (capacitor), resulting in a delayed reaction of the intratracheal pressure to extratracheal pressure changes. The faster the pressure change, the greater the discrepancy between the extratracheal and intratracheal pressure waveforms.

Any cyclic curve can be viewed as a summation of sine waves with various frequencies. The proportion of higher frequencies, also called overtones, is greater in waveforms containing square waves or short, high peaks. The steeper the rising and falling edges of a square wave and the higher and shorter a peak, the greater the high-frequency component contained in it. When such a signal is subjected to low-pass filtering, the high-frequency components are removed first. Incisurae appear in square waves (as in Fig. 4, A and B), because the high-frequency components filling the incisurae have been removed. High, short peaks disappear, steep edges become slanted, but longer peaks and troughs are not equally affected.

In asymmetric oscillation, the shorter inspiratory pressure and flow pulses have a greater proportion of higher frequencies (overtones) than the longer expiratory pulses, and, therefore, are more strongly affected by low-pass filtering. The stronger filtering of the
inspiratory pulses, combined with the better transmis-
sion of the expiratory pulses, results in a lower intra-
tracheal pressure in comparison to the y piece. Therefore, 
asymmetrical oscillation with a shorter inspiration 
than expiration may result in a lower lung volume than 
a constant positive airway pressure of equal mean 
value at the y piece.

With SM50, the inspiratory and expiratory wave-
forms were similar, the oscillation was thus symmetri-
cal. Under this condition, air trapping occurred to a
small degree, as shown by the small but significant 
pressure rise in the airway after damping the endotra-
cheal tube and the intratracheal pressure measure-
ments, which yielded slightly but significantly higher 
mean pressures in the trachea than at the y piece.

Similarly, an increase of lung volume by oscillation 
with symmetrical sinusoidal waves generated by differ-
ent devices was demonstrated in human adults (22), 
isolated dog lungs (8), and neonates with respiratory 
disease (18), in the latter indirectly, with a jacket plethysmograph. Occlusion pressures in normal and 
lung-lavaged rabbits were higher than the respective 
mean pressures at the y piece, when low MAP values 
were used (5). In healthy adult rabbits ventilated with 
SM50 (15), as well as in adult mongrel dogs (24) and 
excised dog lungs (1) ventilated with symmetrical 
 sinusoidal waveforms, intratracheal and alveolar pres-
sures exceeded the pressure at the y piece.

As lung hyperinflation has been noted with different 
HFOV generators and different waveforms (sinusoidal 
and square), and considering the almost-perfect symme-
try of the SM50 waveform with equal inspiratory and 
expiratory amplitude and duration recorded in our 
experiments, the air trapping is unlikely to be caused 
by residual asymmetries in the SM50 waveform. Other 
mechanisms, such as differences between inspiratory 
and expiratory airway impedance, must be involved (8).

It is known from spontaneous breathing as well as from 
conventional ventilation that bronchial elasticity gives 
rise to a higher resistance during expiration than 
during inspiration (21, 28). The extent of this effect in 
HFOV is not known. The abrupt airway diameter 
change at the end of the endotracheal tube may be 
another source of inspiratory-to-expiratory impedance 
differences in higher frequencies. According to an analy-
sis of tube aerodynamics conducted by Bush et al. (7), 
flow from the smaller to the larger diameter (inspira-
tion) faces a smaller impedance than vice versa (expira-
tion), resulting in less inhibition of inspiratory flow 
than expiratory flow and, consequently, leading to air 
trapping. As the total diameter of the bronchial tree 
increases with increasing distance from the glottis, the 
same effect may also play a part in more distal airways. 
These effects can be compensated for or overcome, as 
discussed above, by asymmetrical waveforms with 
shorter inspiration than expiration times.

Air trapping may not affect all parts of the lung 
equally. Marked differences in alveolar pressures were 
noted between different lung lobes of adult rabbits (15), 
isolated fresh and dried pig lungs (27), and a lung 
model consisting of airbags sealed to isolated bronchial