A NEW VALVELESS ALL-PURPOSE VENTILATOR

Description and laboratory evaluation

M. K. CHAKRABARTI AND J. G. WHITWAM

SUMMARY

A ventilator, of new design, is described which has been evaluated on a lung model and in animals. It is simple, versatile, inexpensive and easy to sterilize. A single breathing tube is used in which the respiratory gas is introduced near the patient's airway while a jet in a more distal part of the tube drives the respiratory gas into the patient's lungs. Provided the internal volume between the respiratory gas inlet and the driving jet is at least one tidal volume, the driving gas does not take part in respiratory gas exchange. During controlled ventilation only normal ventilation volumes of respiratory fresh gas are required during both normal and high frequency ventilation. It can be used for any age group with any desired respiratory gas, and is suitable for use in the operating theatre and the intensive care unit. As there are no valves in the breathing system, which is open to the atmosphere at all times, complicated systems for synchronizing the machine with spontaneous breathing are not required. PEEP, NEEP, CPAP and IMV are applied easily.

In recent years a variety of new techniques have been developed to promote appropriate artificial ventilation and weaning from mechanical ventilation: IMV, SIMV, MMV, PEEP, CPAP and high frequency ventilation (HFV) (Hewlett, Platt and Terry, 1977; Lawler and Nunn, 1977; Mushin et al., 1980; Sjostrand, 1980). In seeking to provide many of these facilities in one unit, modern conventional ventilators (Mushin et al., 1980) are becoming increasingly complicated and expensive. However, there has been little change in their breathing systems which still use wide-bore tubing and, since they contain humidifiers in the system, have large compressible volumes. In addition they contain valves which can fail or cause obstruction and, thus, are potentially dangerous. During weaning procedures they require complicated sensing systems. However, apart from their cost and complexity, their main problem is a lack of versatility, particularly in relation to the age of the patient and the frequency of ventilation.

Jet ventilators without large breathing systems have been developed for specific purposes such as high frequency ventilation (Borg et al., 1977; Sjostrand, 1980). These ventilators present problems in delivering a humidified respiratory gas of variable but known composition.

The present paper describes a new, simple, valveless, all-purpose ventilator which overcomes many of the problems associated with the use of existing machines. A subsequent paper will be concerned with its clinical evaluation.

BASIC PRINCIPLE OF VENTILATOR

The principle is to use a single breathing tube in which the respiratory gas is introduced near the patient's airway while a jet in a more distal part of the tube drives the respiratory gas into the lungs (fig. 1). The jet driving gas is independent of the respiratory gas. The distance between the respiratory gas inlet and the jet is sufficient to prevent the driving gas taking part in gas exchange in the lungs. There are no valves or other obstructions in the breathing system, which remains open to atmosphere at all times.

DESCRIPTION OF VENTILATOR

A developed form of the ventilator is illustrated in figure 2.

Breathing system

This consists of a single tube, of any convenient available type, with an internal volume between the driving jet and the respiratory gas inlet of at least one tidal volume, and of sufficient internal diameter to provide a negligible resistance to peak flows. For the adult any conventional, standard disposable tubing with an internal diameter and length of approximately 22 mm and 1.5 m, respectively, is adequate.

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for tidal volumes of up to 500 ml. For children and neonates the tube size can be decreased for convenience, but this is not essential.

**Jets**

The normal driving jet J1 drives the ventilating fresh gas into the patient's lungs. It is designed to generate a maximum pressure in the breathing circuit of 3 kPa (30 cm H₂O) when operated from a source with a pressure of 400 kPa (60 p.s.i.), such as a pipeline.

A second jet J0 (overdrive) can be brought into operation using a special switch S1 to allow the generation of pressures up to 10 kPa (100 cm H₂O) in the breathing circuit for the application of normal tidal volumes at conventional frequencies in patients with very low pulmonary compliance. The switch S1 automatically returns the driving gas from J0 to J1, whenever the ventilator is switched off, so that the high driving pressure can never be applied accidentally to a subsequent patient.

The tidal volume can be varied by adjusting the reducing valve RV1 thereby changing the driving pressure. The driving pressure is indicated by the pressure gauge P1. J2 and J3 can be activated by the switch S2. They provide a continuous flow of gas,

**FIG. 1.** Simplified circuit diagram of the ventilator.

**FIG. 2.** A developed form of the ventilator circuit. J1 = driving jet; J0 = overdrive jet; J2 = PEEP jet; J3 = NEEP jet; RV1 and RV2 = reducing valves; P1 and P2 = pressure gauges; S1 and S2 = switches.
VALVELESS VENTILATOR: DESCRIPTION

thereby generating standing pressures up to 2 kPa (20 cm H₂O), J₂ in the direction of the inspiratory flow and J₃ in the reverse direction to produce PEEP and NEEP, respectively. These standing pressures are indicated by the gauge P₂ and can be regulated by a reducing valve (RV₂).

Conventional high pressure gauges of the type used in this study are still calibrated in p.s.i.; these values were converted to kPa. Both values are quoted throughout.

Respiratory fresh gas (RFG)
The fresh gas is delivered at the connection between the patient’s airway and the breathing system. However, to decrease anatomical deadspace as, for example, during HFV, the respiratory gas may be delivered through a narrow tube placed either in the lumen of an endotracheal tube or in the trachea itself. Only normal respiratory gas flows are required (approximately 100 ml kg⁻¹ min⁻¹). This gas can be from any source (e.g. anaesthetic gases, air, oxygen), it has a constant flow and is warmed and humidified before reaching the patient.

Driving gas
Any driving gas (air, oxygen or nitrogen) from any high pressure source is suitable. To provide the variable frequency of ventilation and variable inspiratory–expiratory time ratios (I:E ratio) a device for “chopping” the driving gas is required and this may be mechanical, electrical or pneumatic.

The consumption of driving gas at any frequency of ventilation is approximately double the ventilation volume because of back flow towards the end of the inspiratory phase. It varies between 10 and 30 litre min⁻¹, depending on tidal volume setting. During PEEP or NEEP another 10–25 litre min⁻¹ gas flow is necessary.

Manual ventilation
Manual ventilation can be instituted either by stopping the jets and intermittently occluding the expiratory port or by occluding the breathing circuit at any other place distal to the RFG input. A variable pressure blow-off valve (3–10 kPa) is incorporated in the breathing circuit for safety (fig. 2). Alternatively, for conventional manual ventilation, a bag with a valve can be connected to the expiratory port, after switching off the jets.

Cleaning and sterilization
The jet manifold is made of metal (brass or steel) which can be autoclaved, and the breathing tube can be made from any conventional ventilator tubing, which is either disposable or capable of sterilization.

Measurement of the volumes of ventilation
Accurate inspiratory and expiratory volumes can be measured by a volume meter or pneumotachograph placed between the airway and the RFG input. However, such a device would add to the deadspace and may not be practical in very small children. The expiratory volume cannot be measured at the expiratory port (exhaust port) because of back flow of the driving jet. However, if a suitable unidirectional volume measuring device (Wright respirometer or pneumotachograph) is placed in the breathing tube between the jet J₁ and the RFG input tube, the inspiratory tidal volume or minute volume will be an underestimate by as much as the volume contributed by the RFG during the expiratory time. However, corrected tidal volume and minute volume can be obtained from a simple nomogram in relation to the RFG flow rate and I:E ratio. Alternatively, the RFG can be measured by an electronic device and added or subtracted from the inspiratory or expiratory volumes, respectively. In this study the tidal volumes were measured using a pneumotachograph system (Gould Medical) with the head placed between the endotracheal tube and the input of the RFG.

Tidal volume and rate controls
The tidal volume can be controlled independently merely by adjusting the jet driving pressure. When the respiratory rate is increased, the inherent property of this ventilator is that the tidal volume decreases. However, the end result of such an increase in rate is to cause a slight increase in ventilation (table II).

Noise
In order to decrease the noise of the jets to acceptable levels, an ordinary bacterial filter (Cape Engineering) is fitted to the expiratory limb of the breathing system. This has the added advantage of ensuring that the expired gas is free of bacteria.

Scavenging
The expired gas can be scavenged with any conventional system.
VENTILATOR PERFORMANCE TEST—MODEL LUNG

This first prototype of the ventilator was tested using lung models in which compliance and airway resistance could be varied to simulate either normal or abnormal lungs as previously described by Chakrabarti and Sykes (1976) and Loh, Sykes and Chakrabarti (1978).

In this prototype a mechanical chopper with a fixed I:E ratio of 1:2 was used. A variable speed electric motor (Citen-Co.) was used to drive a rotating disc, one complete rotation cycle of which represented one respirator cycle, and the driving gas flowed for only one-third of the cycle. The ventilation rate was displayed on a tachometer and the frequency range was 0–200 b.p.m.

RESULTS

Normal frequency ventilation (NFV) 12 b.p.m.

It can be seen in figure 3 that, for any given driving pressure the inspiratory flow rapidly reached a peak and then decelerated to zero as the airway pressure reached its maximum, this being characteristic of any pressure generator (Mushin et al., 1980). With a normal compliance of 50 ml cm H₂O⁻¹ (C₅₀) and a resistance of 5 cm H₂O litre⁻¹ s⁻¹ (R₅) a driving pressure (DP) of 200 kPa (30 p.s.i.) produced a tidal volume (VT) of 0.5 litre (fig. 3). When the airway resistance was increased four times to 20 cm H₂O litre⁻¹ s⁻¹ (R₂₀) there was a slight decrease in tidal volume and DP had to be increased to 207 kPa (31 p.s.i.) to maintain a tidal volume of 0.5 litre. When, in addition, the compliance was decreased to 20 ml cm H₂O⁻¹ (C₂₀) the maximum airway pressure remained constant (at the new driving pressure of 31 p.s.i.) but VT decreased by more than 50%. DP had to be increased to 365 kPa (55 p.s.i.) to restore VT to 0.5 litre and the airway pressure increased to 2.5 kPa (fig. 3).

PEEP and NEEP

The use of jets J₂ and J₃ to provide PEEP and NEEP are illustrated in figures 4 and 5, respectively. It can be seen that this is a relatively "ideal" form of PEEP as there is no retardation of the expiratory flow (fig. 4). In addition, during the application of NEEP (fig. 5) there was virtually no change in the respiratory flow pattern.
VALVELESS VENTILATOR: DESCRIPTION

25

1009

Flow (litre min⁻¹)

Airway pressure (kPa)

Alveolar pressure (kPa)

PEEP

FIG. 4. Lung model. Respiratory gas flow, airway and alveolar pressures during the application of PEEP 1.0 kPa. Ventilation frequency 10 b.p.m.

The range of tidal volumes at 12 b.p.m. which will be provided by this ventilator using the maximum driving pressure without any change in the patient circuit is shown in table I. It can be seen that it is a safe ventilator for any age group since, no matter what the patients condition, the maximum airway pressure which will be developed using jet J1 is 3 kPa (30 cm H₂O).

High frequency ventilation (HFV)

The ventilator was connected to the model lung with normal values for compliance and airway resistance (Cₚ, Rₚ) and VF was increased in stages from 12 to 200 b.p.m. with a constant RFG of 6 litre min⁻¹. It can be seen (fig. 6) that as VF was increased and hence VT decreased the peak airway pressure, peak flow and alveolar pressure decreased, but as the frequency exceeded the response time of the lung, that is greater than 60 b.p.m., a positive end-expiratory pressure developed which was in keeping with previous observations (Chakrabarti and Sykes, 1980). It may be pointed out that there will always be a difference between the alveolar pressure and the pressure measured in the airway because of the pressure gradient resulting from airway resistance and during HFV there is insufficient time for equilibration between these pressures during the respiratory cycle. Hence, the peak inspiratory and end-expiratory pressures will be always higher and lower, respectively, than alveolar pressure.

FIG. 5. Lung model. Respiratory gas flow, airway pressure and alveolar pressure during the application of NEEP. Ventilation frequency 10 b.p.m.
**TABLE 1. Model lung: range of tidal volumes at different lung compliance (C) and airway resistance (R) from adult to neonate using normal driving jet II only (Fig. 2)**

<table>
<thead>
<tr>
<th>Lung compliance (C) (ml cm H₂O⁻¹)</th>
<th>Airway resistance (R) (cm H₂O litre⁻¹ min⁻¹)</th>
<th>Driving pressure (kPa)</th>
<th>Maximum airway pressure (kPa)</th>
<th>Range of Vt (ml)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Adult anaesthetized and paralysed</td>
<td>C₅₀</td>
<td>R₅</td>
<td>0–60 p.s.i.</td>
<td>3.0</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0–400 kPa</td>
<td></td>
</tr>
<tr>
<td>Adult diseased lung</td>
<td>C₂₀</td>
<td>R₂₀</td>
<td>0–60 p.s.i.</td>
<td>3.0</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0–400 kPa</td>
<td></td>
</tr>
<tr>
<td>Paediatric</td>
<td>C₁₀</td>
<td>R₂₀</td>
<td>0–60 p.s.i.</td>
<td>3.0</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0–400 kPa</td>
<td></td>
</tr>
<tr>
<td>Neonate</td>
<td>C₅</td>
<td>R₅₀</td>
<td>0–60 p.s.i.</td>
<td>3.0</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0–400 kPa</td>
<td></td>
</tr>
<tr>
<td></td>
<td>C₁</td>
<td>R₂₀₀</td>
<td>0–60 p.s.i.</td>
<td>3.0</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0–400 kPa</td>
<td></td>
</tr>
</tbody>
</table>

**EVALUATION IN DOGS**

Experiments were performed in five mongrel dogs weighing between 15 and 19 kg. Anaesthesia was induced with methohexitone 10–12 mg kg⁻¹ and maintained with 1% chloralose (initial bolus dose of 3 ml kg⁻¹ followed by an infusion of 1–1.5 ml kg⁻¹ h⁻¹). Suxamethonium 1–2 mg kg⁻¹ h⁻¹ was administered during mechanical ventilation and was

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**FIG. 6. Lung model. Respiratory gas flow, airway and alveolar pressures at different ventilation frequencies (12–200 b.p.m.).**
discontinued to allow the return of spontaneous respiration when required. At the start the dogs were ventilated with a Starling Ideal Pump (conventional) and then transferred to the new ventilator. The left femoral artery and vein were cannulated for measurements of arterial pressure and blood-gas tensions, and the infusion of fluid and drugs, respectively. Arterial blood samples were taken in heparinized plastic syringes for immediate blood-gas analysis. Oxygen was used as the respiratory gas.

A decrease in mean \( P_{\text{ACO}_2} \) from 5.39 to 4.49 kPa and in \( P_{\text{AO}_2} \) from 1.05 to 0.75 kPa (table II, columns 4 and 5). It can also be seen (fig. 7) that a PEEP of approximately 0.15 kPa developed during HFV. A decrease in the driving pressure from 25 to 22 p.s.i. was sufficient to restore \( P_{\text{ACO}_2} \) to 5.2 kPa and the airway pressure decreased even further, to 0.58 kPa (table II, columns 5 and 6).

Where appropriate, statistical analysis was performed using paired \( t \) tests and a probability value <5% was accepted as significant.

### RESULTS

The effects of changing the ventilator (Starling pump to the new machine), doubling the RFG, increasing VF from 12 to 60 b.p.m. and changing the jet driving gas, on airway pressure, \( P_{\text{AO}_2} \) and \( P_{\text{ACO}_2} \) are shown in table II (columns 1, 2, 3 and 4). Changing the jet driving gas from air to nitrogen caused no significant change in \( P_{\text{AO}_2} \) (table II, columns 7 and 8), indicating that the jet driving gas was not participating in gas exchange in the lungs.

### High frequency ventilation (HFV)

Increasing VF from 12 to 60 b.p.m. caused a decrease in mean \( P_{\text{ACO}_2} \) from 5.39 to 4.49 kPa and in airway pressure from 1.05 to 0.75 kPa (table II, columns 4 and 5). It can also be seen (fig. 7) that a PEEP of approximately 0.15 kPa developed during HFV. A decrease in the driving pressure from 25 to 22 p.s.i. was sufficient to restore \( P_{\text{ACO}_2} \) to 5.2 kPa and the airway pressure decreased even further, to 0.58 kPa (table II, columns 5 and 6).

### PEEP and NEEP

The effects of using J2 and J3 to apply PEEP and NEEP respectively are illustrated in figure 8.

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**Table II. Data from five dogs, ventilated with new valveless ventilator. Values refer to mean±SEM. The superscript figures indicate statistically significant differences (P < 0.05) between the relevant columns.**

<table>
<thead>
<tr>
<th>Sample No.</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
<th>6</th>
<th>7</th>
<th>8</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ventilator</td>
<td>Starling pump</td>
<td>NEW</td>
<td>NEW</td>
<td>NEW</td>
<td>NEW</td>
<td>NEW</td>
<td>NEW</td>
<td>NEW</td>
</tr>
<tr>
<td>Respiratory gas</td>
<td>( \text{O}_2 )</td>
<td>( \text{O}_2 )</td>
<td>( \text{O}_2 )</td>
<td>( \text{O}_2 )</td>
<td>( \text{O}_2 )</td>
<td>( \text{O}_2 )</td>
<td>( \text{O}_2 )</td>
<td>( \text{O}_2 )</td>
</tr>
<tr>
<td>Respiratory gas flow</td>
<td>4.0</td>
<td>4.0</td>
<td>8.0</td>
<td>4.0</td>
<td>4.0</td>
<td>4.0</td>
<td>4.0</td>
<td>4.0</td>
</tr>
<tr>
<td>Jet driving pressure (p.s.i.)</td>
<td>—</td>
<td>±3.1</td>
<td>±3.1</td>
<td>±3.1</td>
<td>±3.1</td>
<td>±3.1</td>
<td>±3.1</td>
<td>±3.1</td>
</tr>
<tr>
<td>Jet driving gas</td>
<td>Air</td>
<td>Air</td>
<td>Air</td>
<td>Air</td>
<td>Air</td>
<td>Air</td>
<td>Nitrogen</td>
<td></td>
</tr>
<tr>
<td>Frequency (b.p.m.)</td>
<td>12</td>
<td>12</td>
<td>12</td>
<td>12</td>
<td>12</td>
<td>60</td>
<td>60</td>
<td>12</td>
</tr>
<tr>
<td>( P_{\text{AO}_2} ) (kPa)</td>
<td>58.14</td>
<td>58.43</td>
<td>58.40</td>
<td>58.00</td>
<td>60.39</td>
<td>58.26</td>
<td>59.88</td>
<td>60.04</td>
</tr>
<tr>
<td>±0.54</td>
<td>±1.03</td>
<td>±2.09</td>
<td>±1.54</td>
<td>±1.22</td>
<td>±0.82</td>
<td>±1.06</td>
<td>±1.32</td>
<td></td>
</tr>
<tr>
<td>( P_{\text{ACO}_2} ) (kPa)</td>
<td>5.12</td>
<td>5.05</td>
<td>4.93</td>
<td>5.39</td>
<td>4.49&lt;sup&gt;-6&lt;/sup&gt;</td>
<td>5.20</td>
<td>5.31</td>
<td>5.16</td>
</tr>
<tr>
<td>±0.15</td>
<td>±0.20</td>
<td>±0.26</td>
<td>±0.29</td>
<td>±0.23</td>
<td>±0.28</td>
<td>±0.16</td>
<td>±0.21</td>
<td></td>
</tr>
<tr>
<td>±0.058</td>
<td>±0.050</td>
<td>±0.040</td>
<td>±0.050</td>
<td>±0.011</td>
<td>±0.001</td>
<td>±0.011</td>
<td>±0.010</td>
<td></td>
</tr>
<tr>
<td>Airway pressure (kPa)</td>
<td>1.10</td>
<td>1.05</td>
<td>1.05</td>
<td>1.05</td>
<td>0.75&lt;sup&gt;-6&lt;/sup&gt;</td>
<td>0.85&lt;sup&gt;-6&lt;/sup&gt;</td>
<td>1.00</td>
<td>1.00</td>
</tr>
<tr>
<td>±0.14</td>
<td>±0.07</td>
<td>±0.07</td>
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<td>±0.03</td>
<td>±0.12</td>
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</tr>
<tr>
<td>( \text{FiO}_2 )</td>
<td>5.4</td>
<td>5.45</td>
<td>5.35</td>
<td>5.55</td>
<td>4.50&lt;sup&gt;-6&lt;/sup&gt;</td>
<td>5.68&lt;sup&gt;-5&lt;/sup&gt;</td>
<td>5.50</td>
<td>5.50</td>
</tr>
<tr>
<td>±0.141</td>
<td>±0.07</td>
<td>±0.07</td>
<td>±0.07</td>
<td>±0.07</td>
<td>±0.23</td>
<td>±0.36</td>
<td>±0.28</td>
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<tr>
<td>VT (ml)</td>
<td>282.5</td>
<td>285.00</td>
<td>285.00</td>
<td>265.00</td>
<td>127.50&lt;sup&gt;-6&lt;/sup&gt;</td>
<td>110.00&lt;sup&gt;-6&lt;/sup&gt;</td>
<td>280.00</td>
<td>280.00</td>
</tr>
<tr>
<td>±17.68</td>
<td>±7.01</td>
<td>±7.01</td>
<td>±7.07</td>
<td>±3.54</td>
<td>±4.36</td>
<td>±3.96</td>
<td>±4.35</td>
<td></td>
</tr>
</tbody>
</table>
CPAP

In figure 9 the respiratory flow pattern and airway pressure of a spontaneously respiring dog are shown. The application of 0.8 kPa of CPAP decreased the respiratory rate. The airway pressure was well maintained throughout with an inspiratory dip of only 0.1 kPa.

IMV

In the left-hand side of figure 10, the flow and airway pressure traces are shown in a spontaneously breathing dog with the ventilator running at 12 b.p.m. It can be seen that, during eight inflations by the ventilator, the dog had five spontaneous respiratory movements (that is, its spontaneous rate was between 7 and 8 b.p.m.). Because there are no valves or other obstructions in the ventilator the dog was not "fighting" the machine, and only small changes in the airway pressure were caused by the spontaneous breaths. Following the administration of suxamethonium 2 mg kg⁻¹ the fasciculation caused marked changes in the airway pressure during one inflation of the lungs. Thereafter, the airway pressures and peak gas flows were similar to those during IMV. During IMV the $Fe'CO_2$ was 5.2% and when IPPV was instituted following neuromuscular blockade it increased to 6.3%, showing that the spontaneous respiratory movements had the effect of improving ventilation. Finally, the inflation pressure and tidal volume were increased to return the end-tidal carbon dioxide to 5.0% and the changes required are seen on the right-hand side of the figure.

DISCUSSION

The principle of this new ventilator is to create a pneumatic "piston" from a jet source to drive a respiratory gas into the lungs. It is the first purpose-
built machine which will not only serve as a conventional ventilator, but will also provide high frequency ventilation merely by adjusting the frequency control while still ventilating the patient with any chosen, suitably conditioned gas. The fact that there is as yet no machine commercially available which

FIG. 9. Respiratory gas flow and airway pressure during the application of CPAP 0.8 kPa in a spontaneously breathing dog.

FIG. 10. Respiratory gas flow and airway pressure in a dog. Left side = spontaneous respiration with the ventilator running at 12 b.p.m., that is IMV. $P_{a}^{'CO_2}$ was 5.2% at this stage. Suxamethonium 2 mg kg$^{-1}$ was then administered without any change in the ventilator setting and $P_{a}^{'CO_2}$ increased to 6.3% during IPPV. On the right side, the driving pressure (jet J1, fig. 2) was increased to increase the tidal volume and return $P_{a}^{'CO_2}$ to 5.0%.
will perform this function adequately explains the efforts which are being made to modify existing machines for HFV, for example, the recent modifications of the Nuffield Series 200 Penlon Ventilator by Davey, Lay and Leigh (1982). The idea of using a ventilator (Bird Mk 8) to drive an anaesthetic circuit was described by Voss (1967). However, in such systems the problem of valves remains and they lack a high frequency capability. With the simple chopping device described here only a fixed I:E ratio of 1:2 was available. However, although Borg and colleagues (1977) have recommended an inspiratory time of 22% of the respiratory cycle during HFV, there is as yet little evidence to support this view throughout a whole frequency range. Given a more sophisticated chopping device, this new ventilator will allow more careful investigation of this claim.

When in the ventilation mode, the new ventilator requires only normal flows of respiratory gas, but during spontaneous respiration approximately twice the minute volume is required to ensure carbon dioxide elimination.

Because there are no valves, the ventilated animal does not fight the machine and it provides an excellent IMV system. The ventilator can be run at any frequency and if the minute volume is not adequate to maintain a normal \( P_{\text{CO}_2} \), spontaneous respiration will return which decreases the \( P_{\text{CO}_2} \). Thus, when the volume delivered by the ventilator is progressively decreased an animal will return to normal spontaneous respiration. During this process the RFG should be increased to twice the minute volume to prevent rebreathing.

The machine presents no problems with regard to sterilization since the jet manifold can be autoclaved and the tubing either sterilized or replaced. However, if required, a second bacterial filter can be placed between the manifold and the breathing circuit (fig. 2).

The principal problem with any pressure generator is the variation in ventilation volume as a result of changes in compliance and airway resistance. For this reason flow generators, which can provide a constant expired volume irrespective of such changes, have been preferred for many years. However, even such a conventional ventilator does not deliver constant alveolar ventilation when pulmonary compliance decreases. Although there will be an increase in the airway pressure, measured in the breathing circuit, the expired volume meter will not indicate the decrease in alveolar ventilation since it also records, during expiration, the gas taken up in the compressible volume during inflation. Moreover, the anatomical deadspace may increase because of the high inspiratory pressures which further decreases alveolar ventilation. Thus, the constant volume/flow generators are incapable of maintaining constant alveolar ventilation in the presence of large changes in compliance. A pressure generator immediately responds to changes in lung compliance and, therefore, the measured ventilation volume can indicate deterioration in lung function. This new ventilator, with a simple servo control system to relate the driving pressure to the ventilation volume, has the potential to maintain constant alveolar ventilation. However, such a system would have the same problems as a conventional constant volume machine, that is high inflation pressures with consequent barotrauma and circulatory impairment. Under these conditions it would be more logical to increase the ventilation frequency to maintain alveolar ventilation, a facility which is provided easily by this new design, since a change in HFV causes slight hyperventilation.

In conclusion, a new ventilator is described which, in spite of its simplicity, has remarkable versatility. There are no valves in the breathing circuit so that spontaneous respiration can occur at any time without desynchronization. Because the system is open to the atmosphere at all times, excessive pressure cannot develop and this, together with the absence of valves, makes it very safe. Any respiratory gases, including anaesthetic gases, can be supplied and humidification takes place outside the breathing system. It is suitable for any age group. It can provide PEEP, CPAP, NEEP and IMV. It has the capability for high frequency ventilation while using only normal minute volumes of any chosen respiratory gas. Finally, it is small, simple in operation and maintenance, and inexpensive.

REFERENCES


EIN NEUES KLAPPENLOSES ALLZWECKBEATMUNGSGERÄT
Beschreibung und erprobung im labor

ZUSAMMENFASSUNG

UN NOUVEAU RESPIRATEUR TOUS USAGES SANS VALVE
Description et Etude en Laboratoire

RESUME
Il s'agit de la description d'un respirateur de conception nouvelle qui a été étudié sur un modèle de poumons et chez l'animal. Il est simple, modulable, peu coûteux et facile à stériliser. On utilise un seul tuyau de ventilation dans lequel on administre les gaz inspirés près des voies respiratoires du patient alors qu'un gaz propulseur en amont de ce point entraîne les gaz insérés à l'intérieur des poumons du patient. Dans la mesure où le volume interne entre les gaz inspirés et le gaz propulseur est au moins égal à un volume courant, le gaz propulseur ne participe pas à l'hématose. Au cours de la ventilation contrôlée, seuls des volumes ventilatoires normaux de gaz respiratoires frais sont nécessaires, que la ventilation soit ou non à haute fréquence. Ce respirateur peut être utilisé quelle que soit la classe d'âge et quel que soit le gaz inspiré soi, et il convient aussi bien au bloc opératoire qu'en unité de soins intensifs. Comme il n'y a pas de valves sur le circuit respiratoire qui est ouvert à l'air ambiant en permanence, on n'a pas besoin de systèmes compliqués de synchronisation de la machine et de la ventilation spontanée. Il est facile d'appliquer une PEEP, une NEEP, une CPAP et une IMV.

UN NUEVO VENTILADOR SIN VALVULA PARA CUALQUIER PROPOSITO
Descripción y evaluación en laboratorio

SUMARIO
Se describe un ventilador de diseño nuevo, el cual fue evaluado en un modelo pulmonar y en animales. Es sencillo, versátil, barato y de fácil esterilización. Se usa un tubo de respiración único en el cual se introduce el gas cerca de las vías respiratorias del paciente mientras un chorro en una parte más distante lleva el gas respiratorio dentro de los pulmones del paciente. Siempre que el volumen interno entre la entrada de gas respiratorio y el chorro iguale un volumen respiratorio, el gas no participa del intercambio del gas respiratorio. Durante la ventilación bajo control, se necesitan solamente volúmenes normales de ventilación de gas respiratorio nuevo tanto durante la ventilación normal como la de alta frecuencia. Puede usarse en cualquier grupo de edad con cualquier gas respiratorio deseado y también es adecuado para uso en la sala de operación y la unidad de cuidados intensivos. Puesto que no hay ninguna válvula en el sistema que se encuentra abierto en la atmósfera en todo momento, no se necesitan sistemas complicados de sincronización de la máquina con la respiración espontánea. Se aplican fácilmente los PEEP, NEEP, CPAP e IMV.