THE ASSESSMENT OF VENTILATOR PERFORMANCE

L. LOH, M. K. SYKES AND M. K. CHAKRABARTI

Lung ventilators are important tools in anaesthetic and intensive care practice. They can also be very expensive items. The choice of a suitable ventilator can be difficult, for cheap ventilators have a restricted range of facilities whilst the new generation of ventilators, employing a variety of sophisticated mechanics and electronics, provide a bewildering array of facilities, many of which have not been adequately evaluated in clinical practice. The problem of choice is compounded by the relative lack of useful data on the performance of ventilators. In this article an attempt is made to define the essential features of a ventilator for intensive care use and to introduce the concept of a standard test procedure which could be useful in assessing their efficiency, reliability, safety and suitability for a particular purpose.

**Essential requirements**

1. The ventilator must be safe and reliable.
2. It should be capable of delivering an adequate tidal volume (50-1500 ml) at frequencies from 10 to 50 per minute with an i:E ratio of approximately 1:2 against the range of compliances and resistances encountered in clinical practice.
3. The oxygen concentration of the delivered gas should be known accurately and adjustable over a range of 21-100% oxygen.
4. All controls (tidal volume, frequency, i:E ratio, oxygen concentration, etc.) should be independently variable.
5. The humidifier should be capable of delivering gas with an absolute humidity greater than 32 mg H$_2$O litre$^{-1}$ at the patient Y-piece when the temperature is 37 °C or less. The humidifier should be so designed that the inspired gas temperature can never exceed 41 °C.
6. There should be facilities for continuous monitoring of expired minute volume and airway pressure.
7. There should be provision for adjustable high/low alarms for airway pressure, expired minute volume, and alarms for power source failure and oxygen failure.
8. The gas delivered to the patient must be sterile and a system for ducting away or filtering the expired gas is also desirable.
9. The ventilator breathing system should be either easily sterilized by heat or disposable.

**Other desirable features**

1. Positive end-expiratory pressure (PEEP) of up to 15 cm H$_2$O.
2. Variable i:E ratio.
3. Variable inspiratory flow pattern and end-inspiratory pause.
4. Adjustable positive airway pressure limit from 30 to 100 cm H$_2$O.
5. Facilities for patient triggering, intermittent mandatory ventilation (IMV) and sub-atmospheric expiratory pressure may also be required.
6. Continuous inspired oxygen and expired carbon dioxide monitoring, with adjustable high and low alarms.

Although few ventilators fulfil all the requirements mentioned above, several ventilator systems incorporate many of these features. However, the ventilator specifications provided by the manufacturers seldom give sufficient data on ventilator performance and, understandably, are biased in favour of the ventilator displaying only the desirable features. A major problem is to assess the safety and reliability of a ventilator and whether a ventilator actually performs as it is supposed to.

Recently there has been a document drafted by the International Standards Organization (ISO) committee on breathing machines which has been designed to encourage manufacturers to produce more information about the performance of their machines under standard test conditions. The tests are intended to give objective information on the characteristics of the ventilator, its limitations and its reliability. The results of the test should then be made available to the prospective purchaser to help him in deciding whether the machine can fulfil his particular need. Since the same tests are applied to each machine they...
TABLE I. Endurance test. Ventilator settings and compliance–resistance combinations according to sphere of use

<table>
<thead>
<tr>
<th>Type of ventilator</th>
<th>Tidal volume (ml)</th>
<th>Frequency (b.p.m.)</th>
<th>Compliance (ml cm H₂O⁻¹)</th>
<th>Resistance (cm H₂O.litre⁻¹ s⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Adult</td>
<td>500</td>
<td>20</td>
<td>C₅₀</td>
<td>R₂₀</td>
</tr>
<tr>
<td>Paediatric</td>
<td>150</td>
<td>30</td>
<td>C₂₀</td>
<td>R₅₀</td>
</tr>
<tr>
<td>Neonatal</td>
<td>20</td>
<td>40</td>
<td>C₂</td>
<td>R₂₀</td>
</tr>
</tbody>
</table>

C = compliance; R = resistance. Suffix number indicates value of compliance (ml cm H₂O⁻¹) or resistance (cm H₂O.litre⁻¹ s⁻¹).

will also enable the purchaser to compare the performance of one machine with another. It is also hoped that the tests will encourage the manufacturers to look at their machines critically and stimulate improvements in ventilator design.

The American National Standards Institute* has already published a document, the American National Standard for Breathing Machines for Medical Use (ANS 279.7 1976), which is very similar to the proposed ISO Standard on Breathing Machines. Reference to this document will provide full details of the performance tests and other aspects of ventilator function which we cannot discuss in this article.

The first requirement of any test procedure for a ventilator is to produce a standard lung model. This lung model should, ideally, be easy to construct out of readily available materials, inexpensive, and its characteristics should be easily reproducible. The design of such a lung model has been described by Chakrabarti and Sykes (1976). Essentially, the lung model adopted by the ISO is based on a linear resistance in series with a linear compliance. One of the simplest ways of making such a model is to simulate the lung compliance by a rigid container. This is made preferably of copper or similar material with a high specific heat, and is partially filled with copper wool to absorb the heat of compression of gas during the inspiratory phase, thus rendering it isothermal (Hill and Moore, 1965). The compliance value is dependent on the internal volume of the container. Such a model compliance is easy to construct so that there is a linear relationship between volume injected and pressure change, and a family of compliances of different values can be made to simulate conditions likely to be encountered in clinical practice. A set of model resistances can also be constructed with relative ease (Burton and Fox, 1972; Chakrabarti and Sykes, 1976). Although airway resistance in a patient is not strictly linear, the model resistance is designed to have a linear relationship between flow and pressure difference over a specified range of flows, so that standard characteristics can be easily reproduced in different laboratories.

The standard test utilizes combinations of a compliance and resistance in series to simulate different clinical lung conditions. There are four main tests: an endurance test, a waveform test, a volume performance test and a measurement of internal compliance. These will now be described separately.

The endurance test

A normal production model of the machine is set to run continuously against a lung compliance–resistance combination appropriate to its sphere of use for a period of 2000 h. The conditions of the test are specified in table I. During this test routine maintenance is allowed. The test is designed to assess the reliability of the machine and its components. Following this the machine is subjected to waveform performance tests.

The waveform test

This test enables a functional analysis of the ventilator to be made. The ventilator is connected in turn to each of the compliance–resistance combinations appropriate to its sphere of use in the order shown in table II. The ventilator is initially set to deliver the required tidal volume against the first compliance–resistance combination and a recording of the waveforms is made. The other compliance–resistance combinations are then substituted in turn, initially with no alteration of the ventilator settings, in order to assess the changes in performance with changing lung conditions. Subsequently, the tests are repeated with the ventilator adjusted as necessary to meet the required conditions of the test. Any adjustments necessary should be described. During each part of the test, the waveforms of flow, volume, airway pressure and lung pressure are recorded for examination (fig. 1).
## Table II. Waveform performance test. Showing the compliance–resistance combinations and required ventilation in each of the tests

<table>
<thead>
<tr>
<th>Test sequence</th>
<th>Compliance (ml cm H$_2$O)</th>
<th>Resistance (cm H$_2$O.litre$^{-1}$s$^{-1}$)</th>
<th>Tidal volume (ml)</th>
<th>Frequency (b.p.m.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Adult</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>C$_{50}$</td>
<td>R$_{5}$</td>
<td>500</td>
<td>20</td>
</tr>
<tr>
<td>2</td>
<td>C$_{50}$</td>
<td>R$_{50}$</td>
<td>500</td>
<td>20</td>
</tr>
<tr>
<td>3</td>
<td>C$_{20}$</td>
<td>R$_{20}$</td>
<td>500</td>
<td>20</td>
</tr>
<tr>
<td>4</td>
<td>C$_{20}$</td>
<td>R$_{20}$</td>
<td>500</td>
<td>20</td>
</tr>
<tr>
<td>Paediatric</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>C$_{20}$</td>
<td>R$_{20}$</td>
<td>300</td>
<td>20</td>
</tr>
<tr>
<td>2</td>
<td>C$_{20}$</td>
<td>R$_{20}$</td>
<td>300</td>
<td>20</td>
</tr>
<tr>
<td>3</td>
<td>C$_{10}$</td>
<td>R$_{10}$</td>
<td>300</td>
<td>20</td>
</tr>
<tr>
<td>4</td>
<td>C$_{10}$</td>
<td>R$_{10}$</td>
<td>300</td>
<td>20</td>
</tr>
<tr>
<td>5</td>
<td>C$_{2}$</td>
<td>R$_{2}$</td>
<td>50</td>
<td>30</td>
</tr>
<tr>
<td>6</td>
<td>C$_{2}$</td>
<td>R$_{2}$</td>
<td>50</td>
<td>30</td>
</tr>
<tr>
<td>7</td>
<td>C$_{2}$</td>
<td>R$_{200}$</td>
<td>50</td>
<td>30</td>
</tr>
<tr>
<td>Neonatal</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>C$_{3}$</td>
<td>R$_{50}$</td>
<td>30</td>
<td>30</td>
</tr>
<tr>
<td>2</td>
<td>C$_{3}$</td>
<td>R$_{50}$</td>
<td>30</td>
<td>30</td>
</tr>
<tr>
<td>3</td>
<td>C$_{1}$</td>
<td>R$_{200}$</td>
<td>30</td>
<td>30</td>
</tr>
<tr>
<td>4</td>
<td>C$_{1}$</td>
<td>R$_{200}$</td>
<td>30</td>
<td>30</td>
</tr>
<tr>
<td>5</td>
<td>C$_{1}$</td>
<td>R$_{1000}$</td>
<td>30</td>
<td>30</td>
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<tr>
<td>6</td>
<td>C$_{1}$</td>
<td>R$_{1000}$</td>
<td>15</td>
<td>60</td>
</tr>
<tr>
<td>7</td>
<td>C$_{1}$</td>
<td>R$_{1000}$</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

1 : E ratio as close to 1 : 2 as possible. Notation as in Table I.

Comparison of the flow patterns with changing lung characteristics will determine whether or not the machine should be classified as a flow or pressure generator. Examination of the volume trace will show if the ventilator is capable of delivering the required ventilation under the conditions simulated by the test and also will define the conditions under which the machine is likely to fail. Other features such as leakage through incompetent mechanical valves and design faults may also be revealed by the waveform test.

### Volume performance test

This test is designed to show the maximum volumes of gas which can be delivered out of the machine over a wide range of frequencies when the 1 : E ratio is 1 : 2 and the compliance–resistance combination is appropriate to the ventilator’s sphere of use (table III). Some machines are unable to deliver a small enough tidal volume to satisfy paediatric or neonatal patients, while in other machines the maximum tidal volume decreases at high frequencies.

### The internal compliance of a ventilator

This is a measure of the volume of gas compressed in the ventilator, humidifier and conducting tubing.

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**Fig. 1.** Diagram of apparatus to record the waveform performance test.
**Table III. Volume performance test. Required compliance–resistance combinations and frequencies appropriate to the sphere of use of the ventilator**

<table>
<thead>
<tr>
<th>Type of ventilator</th>
<th>Compliance (ml cm H₂O⁻¹)</th>
<th>Resistance (cm H₂O.litre⁻¹ s⁻¹)</th>
<th>Frequency (b.p.m.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Adult</td>
<td>C₂₀</td>
<td>R₂₀</td>
<td>10, 15, 20, 30</td>
</tr>
<tr>
<td>Paediatric</td>
<td>C₁₀</td>
<td>R₅₀</td>
<td>15, 20, 30, 40</td>
</tr>
<tr>
<td>Neonatal</td>
<td>C₅</td>
<td>R₂₀₀</td>
<td>15, 20, 30, 40</td>
</tr>
<tr>
<td></td>
<td></td>
<td>R₂₀₀</td>
<td>20, 30, 40, 60</td>
</tr>
</tbody>
</table>

Notation as in table I.

per unit change in airway pressure. The gas volume compressed in the patient tubing is not delivered to the patient but is included, together with the gas from the patient, in the measured expired minute volume. The volume of gas lost to the patient as a result of compression may become a significant proportion of the volume output of the ventilator if airway pressures are high and, in particular, if tidal volumes are small as in paediatric and neonatal ventilation.

**INVESTIGATION OF A VENTILATOR**

The performance of the Bennett MA 1B ventilator was investigated on a lung model constructed according to ISO specifications. The results of the test will be used to illustrate in more detail some of the points mentioned above. Firstly, a brief description of the working principle of the Bennett MA 1B ventilator is required. The ventilator is essentially a “bellows in a bottle” arrangement. A flow of air from an electrically powered compressor passes through the main solenoid valve to drive a venturi system producing a high flow of air. The gas passes via a peak flow control valve into a chamber containing a concertina bellows (fig. 2). The bellows is thus compressed and gas contained in the bellows is delivered to the patient through the inspiratory valve while the expiratory valve is closed. The volume delivered from the bellows is dependent on the setting of a volume control potentiometer. The base plate of the bellows is linked to a similar potentiometer and when the electrical output of this matches the electrical output of the volume control potentiometer, the electrical logic system of the ventilator closes the main solenoid valve, thus terminating the inspiratory flow, and the machine cycles to expiration.

We will now examine in detail some of the traces obtained in a waveform test performed on this ventilator. The ventilator controls were unaltered in all the traces and only compliance–resistance combinations were changed. Figure 3 shows waveforms of flow, volume, airway pressure and lung pressure derived from the lung model with different compliance–resistance combinations. Considering the flow waveform first, it will be seen that when using the same compliance and increasing the airways resistance, the peak flow decreases progressively. The shape of the waveform alters from one which approximates to a square wave to one with a more curved shape and a slower increase in flow. Not only does the shape of the flow wave change, but the duration of inspiration also increases slightly with increasing resistance. Note also the decrease in flow which occurs when compliance is decreased. Since the flow pattern is altered by changes in compliance and resistance, the machine is behaving as a pressure generator during inspiration. This is to be expected, as the flow through a venturi system is, to a degree, dependent on the down-stream pressure. The expiratory flow also alters with the compliance and resistance combinations, the product of the two (the
time constant) determining the pattern of the exponential flow out of the lung model.

The volume trace shows a decreasing volume delivered to the lung with increasing resistance and decreasing compliance, and this is related to increasing airway pressure. The increase in the volume of gas compressed in the ventilator system during inspiration with increasing airway pressure can almost wholly account for the decrease in delivered volume to the lung. This will be discussed later. However, it must be pointed out that the required volume can be achieved by further adjustment of the machine and this is clearly seen in the volume performance test.

The change in lung pressure during inspiration decreases with increasing resistance reflecting the smaller volumes delivered to the lung, but, in addition, it can be seen that with the greater airway resistances a standing positive pressure is developed in the lungs. This is a result of the longer time constant of the compliance–resistance combination retarding expiratory flow to such an extent that the lung does not empty fully before the onset of the next inspiratory phase.

The volume performance test

Figure 4 shows that the minimum tidal volume remains relatively constant at all frequencies and the machine is satisfactory for paediatric patients. However, with the i : e ratio set at 1 : 2 throughout, the maximum volume decreases with increasing frequency; but the volume only decreases significantly at volume–frequency combinations which are seldom used clinically, and the machine is capable of delivering a more than adequate volume of ventilation under most clinical circumstances. The decrease in volume at high frequencies is caused by the high inspiratory flows producing airway pressures equal to the maximum safety pressure of the machine and causing gas to blow off to atmosphere. It is quite probable that much larger volumes could have been delivered at the higher frequencies if the compliance of the lung model was increased and the resistance decreased so that blow-off pressures would not be reached. This is a serious criticism of the test, since, as it stands, the test only shows when the pressure blow-off comes into operation and does not reveal the true maximum volume output of the machine.

Internal compliance

The internal compliance test was performed using the method described by Loh and Chakrabarti (1971). A pressure-operated collect valve (Sykes, 1969) separated the volume of gas discharged from the lung
Figure 4. Volume performance test of Bennett MA 1B. Showing range of volumes (maximum and minimum) which can be achieved at the stated frequencies against a C\textsubscript{20} R\textsubscript{20} lung model at an 1 : E ratio of 1 : 2.

during expiration from that compressed in the tubing compartment. Swept gas volume (the volume of gas delivered from the machine with no impedance to flow) was also measured in order to calculate the internal compliance of the bellows compartment.

\[
\text{Internal compliance of tubing compartment} = \frac{V_{CT}}{\text{End-inspiratory pressure}}
\]

\[
\text{Internal compliance of bellows compartment} = \frac{(V_S) - (V_L + V_{CT})}{\text{End-inspiratory pressure}}
\]

where

\(V_S\) = swept gas volume

\(V_L\) = volume of gas from lung

\(V_{CT}\) = volume of gas from tubing compartment

As mentioned earlier, the tidal volume is determined by the volume control setting and the concertina bellows is compressed from the full position to a point which is determined by the set volume. It follows that the smaller the tidal volume displaced the larger the residual volume in the concertina bellows, and the more gas will be compressed into this residual volume at any particular pressure. The internal compliance of the bellows compartment was therefore measured at different tidal volumes.

Figure 5 shows graphically the internal compliance values of the Bennett MA 1B at different volume settings using standard Bennett tubing, but without a humidifier. The value for the tubing compartment remains relatively unchanged with different tidal volumes, but the bellows internal compliance decreases with increasing tidal volume as expected.

The waveform test illustrated above was performed with a tidal volume setting of 500 ml which would give an internal compliance of 2.5 ml cm H\textsubscript{2}O\textsuperscript{-1} and 3.2 ml cm H\textsubscript{2}O\textsuperscript{-1} for the bellows compartment and tubing compartment respectively and a total internal compliance of 5.7 ml cm H\textsubscript{2}O\textsuperscript{-1}. Examination of the volume and airway pressure traces will show a difference in volume delivered of about 100 ml between C\textsubscript{50} and R\textsubscript{5} and C\textsubscript{20} and R\textsubscript{5} combinations and a difference in peak airway pressure of 22 cm H\textsubscript{2}O. This would give a predicted decrease in tidal volume delivered to the lung of 125 ml. The changes in volume delivered by the ventilator with the various lung compliance–resistance combinations can therefore be accounted for to a large extent by the differences in the volume of compressed gas at the different airway pressures.

We have presented only a portion of the information derived from the testing of the Bennett MA 1B on the ISO lung model system. Our aim is to illustrate the kind of insight into the nature of the machine which results from such tests. We have also looked at several other ventilators in the same way and have been impressed by the frequency with which major faults in ventilator design have been revealed and
ventilator malfunctions have been picked up. We shall briefly describe three examples:

A volume performance test was performed on ventilator “A”, which revealed a marked decrease in the volume output with increasing frequency (fig. 6). In order to investigate this further the ventilator was tested with a volume setting of 500 ml and i:E ratio of 1:2 at frequencies from 10 to 50 per minute (fig. 7). Even under these circumstances the volume output still decreased with increasing frequency. With the frequency fixed at 20 per minute the volume output was then measured at different i:E ratios (fig. 8). The volume appeared to decrease with shortening expiratory time. It was concluded that the fault lay in the relatively slow filling of the inspiratory bellows under gravity because of the high resistance to flow of the intake valve and conducting tubing of the bellows. The higher frequencies shortened the expiratory phase time, limited the filling of the inspiratory bellows and thus decreased the volume delivered from the bellows during inspiration. This was brought to the notice of the manufacturers, who have attempted to improve the resistance characteristics of the bellows intake valve system.

A simple ventilator intended for anaesthetic use, ventilator “B”, was observed to produce less chest movement than would have been expected from the expired tidal volume measurement. When tested on the lung model, it was found that increasing the inspiratory flow setting caused a decrease in tidal volume delivered to the lung. A spirometer situated at the expiratory port showed that there was a leak of gas through the expiratory valve at the beginning of the inspiratory phase. The leak varied from 100 ml per breath at the minimum flow setting to 270 ml per breath at the maximum flow setting. The nature of the fault had not been immediately recognized when the machine was in clinical use because the expired minute volume had been monitored using a new electronic Wright’s respirometer (BOC Medi-shield Ltd). The expired minute volume on this equalled the fresh gas flow, but did so because it included the leaked gas volumes. Had the older type of respirometer been used, the fault would have been picked up immediately since the meter would have displayed volume change early in inspiration as well as during expiration. A second new machine of the same type was similarly tested and found to have the same fault. Many others also had been in use in clinical practice without the fault being recognized. The manufacturers were informed and the pressure-operated expiratory valve has since been modified to diminish the inspiratory leak.

Ventilator “C” was obtained from the manufacturers for evaluation. A waveform test was performed and was shown to give satisfactory results. However, the expiratory flow pattern was not exponential and subsequent testing revealed that there was no gas issuing from the expiratory port during expiration. Further examination of the machine revealed that the inspiratory non-return valve was missing. Gas which had been delivered to the lung model was re-entering and filling the inspiratory bellows during expiration because of the absence of the valve. Had the machine been attached to a patient, serious rebreathing would have occurred unless the expired volume had been monitored. We would strongly recommend that the expired tidal volume be measured as well as the volume delivered to the lung during ventilator performance tests in order to pick up faults in the inspiratory and expiratory valves.

Fig. 6. Volume performance test of ventilator “A”. Performed with different compliance–resistance combinations. Showing decreasing maximum volume output with increasing frequency.
Fig. 7. Traces of ventilator "A" showing progressive decrease in delivered tidal volume with increasing frequency. Volume setting unchanged and I:E ratio 1:2.

Fig. 8. Traces of ventilator "A" showing progressive decrease in delivered tidal volume with shortening expiratory time. Volume setting unchanged and frequency 20 b.p.m.
The preceding examples serve to stress the importance of ventilator testing. Although we have singled out three examples only, we would like to point out that very few of the ventilators we have evaluated were without fault. Although the manufacturers have all been extremely co-operative and keen to improve their ventilators, it is apparent that some of the faults were errors in design which should have been corrected before the machines were put on the market. The ISO performance testing procedure is a major step towards a more rational approach to ventilator assessment. While the tests may require modification and elaboration in the future, it is hoped this more objective and critical approach to ventilator performance will lead to the design of better and safer ventilators.

The performance tests should be used by manufacturers to test their machines at the design stage to sort out problems before the models are in production; tests could also be employed on the production line for the purposes of quality control. It is also important that ventilators are assessed independently by anaesthetists and perhaps a centre run by anaesthetists to test new models could usefully act as a consumer guide. We would also like to stress the need for anaesthetists to test their own ventilators in order to check that they are functioning satisfactorily. A relatively inexpensive model lung such as the Manley Lung Ventilator Performance Analyser (BOC Medishield Ltd) could be used very effectively for this purpose (Chakrabarti and Sykes, 1976).

The tests discussed in this article are only the first of a range of procedures which could be used to evaluate many aspects of ventilator performance, for example, the accuracy of oxygen mixing apparatus and other controls and monitors, the efficiency of humidification, patient triggering devices, PEEP, IMV, etc. Such tests should be constantly under review to keep pace with advances in ventilator design.

REFERENCES