SOME INFANT VENTILATOR SYSTEMS
A Description of their Characteristics and Function

BY

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SUMMARY
The pressure:flow relationships which occur when four infant ventilator systems are attached to model lungs have been investigated. The results show that all four systems were capable of ventilating model lungs with satisfactory values for minute volume but two of the systems had features due to poor expiratory pathway design that were considered unsatisfactory. The expiratory pathway of a T-piece occluder showed considerable advantage and was combined with an improved method of fresh gas flow delivery in a new system. These findings are discussed and in view of the importance of the characteristics of the driving ventilator in interpreting these findings, the time relationships which occur in the cycling of the Newcastle ventilator have been investigated and are included as an appendix.

During the past decade the use of nitrous oxide, oxygen and muscle relaxants with intermittent positive pressure manual ventilation has become commonplace in paediatric anaesthesia. Most anaesthetists now agree that in adult practice controlled ventilation is best performed by a ventilator, rather than by hand, for the reasons summarized by Mushin, Rendell-Baker and Thompson (1959). The same reasons apply even more forcibly in paediatric anaesthesia, provided that a ventilator is available which is flexible enough in its function to work over the wide range of frequency and tidal volume which may be required and in the face of varied conditions of airway resistance and compliance.

The problems of providing automatic ventilation in infants and children have been reviewed by Mushin, Mapleson and Lunn (1962) who deduced criteria that must be fulfilled by a ventilator if its function is to be satisfactory under these circumstances. Detailed information regarding the mode of function of infant ventilator attachments is not easily available. The studies which follow are part of the work undertaken when we set out to find an improved replacement for the attachment in current use, and they provide an analysis of the mode of function of the attachment and that of three others which we chose as alternatives.

METHODS
Model lungs were constructed with values for compliance and airway resistance near to those of a neonate and a 1-year-old child. The values used are similar to those given by Mushin, Mapleson and Lunn (1962). Lung compliance was represented by compression of air within a glass bottle whilst airway and endotracheal tube resistance were represented by tubing of a suitable bore inserted into the bottle. The values that actually existed for compliance and airway resistance are shown in table I. In these model lungs there was a linear relationship between pressure rise and tidal volume over the range of tidal volumes used.

### Table I

<table>
<thead>
<tr>
<th>Type of Lung</th>
<th>Compliance (ml/cm H₂O)</th>
<th>Airway resistance cm H₂O/(10 l./min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Neonate model lung</td>
<td>4.1</td>
<td>18</td>
</tr>
<tr>
<td>1-year-old model lung</td>
<td>7.7</td>
<td>8</td>
</tr>
</tbody>
</table>

Measurement of gas flow at the junction of the airway resistance and attachment of the circuit being investigated was made using a modified pneumotachograph head and a differential capacitance manometer. Integrated readings from the
manometer were used to measure the expired minute volume (Lunn, Molyneux and Pask, 1965). Pressures at various points in the ventilator circuit and anaesthetic circuit were measured using a Greer (1958) defocusing manometer. Flow and pressure recordings were displayed on a Southern Instruments ultra-violet recorder. The recordings chosen to illustrate this paper are reproductions of tracings from the originals and are presented in this way for the sake of clarity.

THE DRIVING VENTILATOR

The ventilator circuits A, B and D are attachments driven by the most recent Newcastle ventilator (Burn, 1967). This is a pressure-cycled flow generator and is the ventilator from which the Cyclator was developed by the British Oxygen Company. It differs from the commercial version in that the expiratory pathway is separate and includes a power-operated expiratory valve. This expiratory valve closes coincidentally with the start of the inspiratory phase and opens widely to atmosphere as soon as inspiration is completed. It is an essential feature of the driving ventilator when circuit D is used.

Circuit A.

This circuit provided the original, interim solution to our problem of automatic ventilation in infants and children during anaesthesia. It is based on the principle that since manual compression of an open-ended bag attached to the expiratory limb of a T-piece circuit had been shown to be satisfactory in theory and in practice (Harrison, 1964a; Rees, 1950; Nightingale, Richards and Glass, 1965) the automatic operation of the same system would be the logical first step in providing a solution (Inkster and Lunn, 1964).

The circuit is shown in figure 1. An open-ended bag attached to the expiratory limb of a T-piece is contained in a glass jar of approximately 1000 ml capacity. This bag is compressed within the bottle by inflating pulses from the driving ventilator. When faced with the reduced lung compliance and increased airway resistance of an infant, the ventilator will reach its cycling pressure rapidly without significant flow occurring. The circuit has been modified by the inclusion on the inflation line of a latex bag which acts as an in-parallel compliance. The extent to which additional compliance is included in the circuit can be varied by the adjustment of a screw clip on the tube leading to the latex bag. In this way the ventilator faces conditions of compliance and airway resistance nearer to those of the adult, for which it was designed, and pressure cycling occurs only after significant flow has been achieved. This modification is similar to the "dummy lung" referred to by Mushin, Mapleson and Lunn (1962).

The open-ended bag is not an essential feature since it is in free communication with the bottle. It is retained because movement of the bag gives a rough indication of effective function.

We believe that the system operates in the following manner.

A positive pressure pulse from the driving ventilator compresses gas within the bottle. The fresh gas flow at the T-piece is directed into the lungs and is supplemented by a variable volume
of gas from the expiratory limb of the T-piece. The volume of this combined inflation depends on the length of time taken for the pressure to rise in the bottle and so to cycle the driving ventilator. The duration of the inspiratory phase at a given pressure setting of the ventilator depends on the degree of inclusion of the in-parallel compliance in the circuit. Once the cycling pressure is achieved the expiratory pathway of the ventilator opens and expiration takes place via the expiratory limb of the T-piece, the reservoir bag, the bottle, and finally through the expiratory valve of the driving ventilator.

Typical records of inspiratory and expiratory flow with airway, lung and bottle pressures are shown in figure 2. Arbitrary division of inspiration into phases has been made to clarify description of events.

During phase 1 there is a rapid rise in inspiratory flow rate and airway pressure as the inflation pulse from the driving ventilator increases pressure in the bottle. The inspiratory flow then shows a brief fall, at the end of phase 1, as filling of the in-parallel compliance takes place.

During phase 2 the inspiratory flow rate and the airway pressure continue to rise, but at a slower rate because the in-parallel compliance is distending and "damping" the rise in bottle pressure. During this time the lung pressure rises as the lungs are filled by gas diverted from the fresh gas inflow of the T-piece and by gas derived from compression of the bag on the expiratory limb. Peak inspiratory flow and airway pressure are reached and the ventilator cycles.

During phase 3, pressure within the bottle falls at a slow rate because of the poor outflow design, which is largely due to gas from the in-parallel compliance discharging into the same outflow path. All the fresh gas cannot escape down the expiratory limb until the pressure in the bottle has fallen to a critical level. Until this happens a decreasing proportion of the fresh gas flow will still be diverted to the lungs. This is the phase of "split gas flow" and is seen as a decreasing inspiratory flow rate with a slowly increasing lung pressure.

Phase 4, expiration, begins when the bottle pressure has fallen sufficiently to allow all the fresh
gas flow to pass down the expiratory limb, i.e. the airway pressure is higher than the bottle pressure. Lung pressure begins to fall. The expiratory flow rate does not reach high peak values and continues at a slowly decreasing rate until the next inspiration begins. The fall in airway and lung pressures is slow and, even with a significant expiratory-inspiratory pause, atmospheric pressure is not reached before the start of the next inspiration.

Although this circuit has been satisfactory in clinical use and produced adequate ventilation in model lungs with the production of values shown in table II, several comments can be made. The in-parallel compliance is not effective at the start of the inspiratory phase. Inspiration is completed by a "split fresh gas flow phase" and the extent to which this plays a part depends on the fresh gas flow rate. During the expiratory pause, residual flow, lung and airway pressures remain at the start of the next inspiration. These last features are due to turbulence at the T-piece, resistance at the tail of the bag, emptying of the in-parallel compliance and the resistance of the expiratory valve of the driving ventilator.

These data led us to investigate, in a similar fashion, other circuits as possible alternatives.

**Circuit B.**

This circuit (fig. 3) is the commercially available Blease Paediatric Ventilator attachment and consists of a closed bag of 1,000 ml capacity contained in a bottle of 3,000 ml capacity. The driving ventilator is attached to the bottle, the inflation and expiratory lines ending in a common port. This attachment is designed to be operated by the Deansway Pulmoflator but we have used the Newcastle ventilator to drive the attachment in the belief that most anaesthetists will use whatever ventilator seems suitable to operate it. We are aware that some of the observations made may reflect the particular characteristics of the Newcastle ventilator and best performance may only be obtained by using the Pulmoflator.

This circuit would be expected to operate in this manner. Fresh gases from the anaesthetic machine are diverted into the closed bag by a lightly loaded diversion valve. The inflation pulse

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**TABLE II**

<table>
<thead>
<tr>
<th>Circuit</th>
<th>Model lung</th>
<th>Frequency (f)</th>
<th>Vt (ml)</th>
<th>Fresh gas inflow (ml/min)</th>
<th>Pressures (cm H$_2$O)</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>Neonate</td>
<td>42</td>
<td>26</td>
<td>8000</td>
<td>+1.0 to 7.5</td>
</tr>
<tr>
<td></td>
<td>1-year-old</td>
<td>28</td>
<td>58</td>
<td>8000</td>
<td>+1.0 to 7.5</td>
</tr>
<tr>
<td>B</td>
<td>Neonate</td>
<td>34</td>
<td>29</td>
<td>1000</td>
<td>+0.5 to 7</td>
</tr>
<tr>
<td></td>
<td>1-year-old</td>
<td>27</td>
<td>56</td>
<td>1500</td>
<td>+0.5 to 7</td>
</tr>
<tr>
<td>C</td>
<td>Neonate</td>
<td>37</td>
<td>24</td>
<td>6000</td>
<td>-1.0 to 7</td>
</tr>
<tr>
<td></td>
<td>1-year-old</td>
<td>30</td>
<td>57</td>
<td>10000</td>
<td>-2.0 to 7</td>
</tr>
<tr>
<td>D</td>
<td>Neonate</td>
<td>42</td>
<td>26.5</td>
<td>4000</td>
<td>0 to 7</td>
</tr>
<tr>
<td></td>
<td>1-year-old</td>
<td>32</td>
<td>47</td>
<td>8000</td>
<td>0 to 6.5</td>
</tr>
</tbody>
</table>

* In the case of circuit C the bottle pressures refer to the weight-loading on the flap operating the reed switch.
from the driving ventilator raises the pressure within the bottle and closes the valve on the expiratory limb. At the same time the bag is compressed and fresh gas is expelled from it to the patient. When the driving ventilator cycles, the bottle pressure falls and releases the expiratory valve. This allows expiration to take place via the expiratory limb, expiratory valve, the bottle, and the expiratory line and valve of the driving ventilator to atmosphere.

Experimentally we have used this device as a volume-cycled ventilator attachment as recommended by the manufacturers. Fresh gas flow passing to the circuit was adjusted so that, with every inflation pulse from the driving ventilator, the bag within the bottle collapsed completely. The pressure developed by the driving ventilator in the bottle was adjusted to exceed peak airway pressure.

Traces of typical flow and pressure events are reproduced in figure 4. At the start of inspiration in phase 1 a rapid rise in pressure within the bottle causes a rapid rise in inspiratory flow rate and airway pressure with a slower rise in lung pressure. Peak inspiratory flow and pressure values are reached as the bag empties. During phase 2 the inspiratory flow and airway pressure fall as the final content of the bag is expelled and lung pressure rises slowly. This is associated with delay in the attainment of the driving ventilator cycling pressure caused by the need to compress a large volume of air within the bottle. When the driving ventilator cycles, the bottle pressure begins to fall. The expiratory valve is maintained in the closed position until the bottle pressure has fallen to a value less than the pressure in the expiratory limb. During the latter part of phase 2 there is therefore a closed outflow path. Continuing fresh gas inflow results in a slow continued rise in lung pressure and in figure 4 the inspira-
tory flow is seen to be maintained at a rate near to the fresh gas inflow. Since very high pressure may be required to squeeze the last few ml out of a rubber bag and because at this stage the pressure in the bottle is being relaxed, the first few ml may enter the bag and the inspiratory flow will be shown as less than the fresh gas inflow (fig. 4A). The inspiratory flow pattern is thus biphasic, consisting of (a) expulsion of bag contents and an element of fresh gas flow and (b) the fresh gas inflow alone (a T-piece occlusion effect).

During phase 3 expiratory flow is at first rapid but later slows because both the expired gas and the gas compressed within the bottle must exhaust through a common path owing to the normal exponential decay. Sustained airway and lung pressures occur and there is a residual pressure of 0.5 cm H₂O during the expiratory: inspiratory pause due to an outflow design which incorporates a weight-loaded valve.

If the device is used as described it is theoretically possible to calculate the minute volume. Unless there is a leak in the circuit all the measured fresh gas flow passes via the reservoir bag or directly into the lungs. Using an integrating pneumotachograph we have confirmed that under these circumstances the fresh gas flow does equal the minute volume. We found that volume-cycled operation was difficult in clinical practice since the low fresh gas flows which will be required, e.g. 600 ml/min, are difficult to measure to an accurate degree with conventional anaesthetic flowmeters. This is particularly so when mixtures of nitrous oxide and oxygen are to be given, and the vaporization of accurate percentages of halothane at these low flow rates presents problems.

The effect of increasing the fresh gas flow by serial increments has been investigated (fig. 5). Without any alteration of the pressure or frequency settings of the driving ventilator, the reservoir bag will, at first, continue to expel its contents completely, causing a progressive rise in the peak values for inspiratory flow, airway pressure and lung pressure. There is also an increase in the residual pressure in the system during the expiratory: inspiratory pause. The circuit continues to act as a volume-cycling device but there is progressive shortening of the T-piece occlusion factor of the inspiratory flow pattern. The rising airway pressure is approaching the value of the bottle pressure and there is therefore less delay in opening of the expiratory valve. When peak airway and peak bottle pressures are nearly equal the expiratory valve opens coincidentally with cycling of the ventilator but the contents of the bag may not be fully expelled, i.e. pressure cycling begins. With still further increase in fresh gas flow the bag will not collapse completely at each inflation phase and soon becomes fully distended. The excess gas "spills" over by lifting the diversion valve during the expiratory pause and this is shown as "chatter" on the expiratory flow tracing. This excess gas escapes along the expiratory pathway and can cause an increase in resistance to outflow. It is important to realize that this circuit can operate in these two distinct ways.

Using this circuit we were able to ventilate the
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model lungs with adequate tidal volumes and pressures, the values of which are shown in table II.

Circuit C.

This circuit is a T-piece occluder, of the type first described by Keuskamp (1963) and later by Reynolds (1964). Commercial versions are now available but the circuit used in this investigation was developed in our laboratory and is shown diagrammatically in figure 6. A brass weight seats on to the end of the expiratory limb of the T-piece, diverting the fresh gas into the lungs. The duration of inflation is determined by the rate at which the fresh gas inflow can develop sufficient pressure in the airway to raise a weighted flap. The device is thus pressure-cycled since upward movement of the flap activates a switch which completes the circuit to a solenoid. The weight is raised from the end of the T-piece by the action of the solenoid and expiration takes place to atmosphere. The time interval between raising of the occluding weight and its fall, when another inspiration is initiated, is controlled by an electronic timing circuit. Adjustment of the fresh gas flow will control the duration of inspiration and manipulation of the rate control will permit variation of the inspiratory : expiratory ratio. This ratio and the rate are displayed on meters and enable calculations of tidal volumes to be made, e.g.:

- Fresh gas flow rate 3000 ml/min.
- Inspiratory : expiratory ratio 1 : 2.
- Respiratory rate 40/min.

Fresh gas flow rate during the inspiratory phase 1000 ml.
Tidal volume 1000/40 = 25 ml.

The fresh gas inlet in this device is directed towards the expiratory limb of the T-piece and acts as a negative pressure generator during expiration. The maximum negative pressure developed, if the fresh gas flow does not exceed 10 litres/min, is —2 cm H2O.

Reproductions of typical traces of inspiratory and expiratory flow, lung and airway pressures are shown in figure 7. At the start of inspiration, the inspiratory flow rate rises almost simultaneously to a flow rate equal to the fresh gas flow rate. This is associated with a steep rise in airway pressure and commencing rise in lung pressure. The inspiratory flow rate is sustained as the airway and lung pressures reach maximum, causing cycling of the device. The weight is raised from the end of the T-piece and a precipitous fall in airway pressure ensues which finally becomes subatmospheric owing to the reverse fresh gas inflow. Expiratory flow rate is rapid and completed well before the start of the next inspiratory phase. Lung pressure falls rapidly to subatmospheric levels.

The values for tidal volume and the pressures achieved when this system was used to ventilate model lungs are shown in table II. In clinical use we find that it is difficult to adjust the interrelated controls of fresh gas inflow and rate so that the desired ventilation is achieved. In its present form there is no control over the character of the in-

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spiratory flow pattern and there is no means of producing a slow onset to inspiration if this is thought to be desirable clinically. A major theoretical advantage is the ability to calculate the tidal volume but unfortunately the accuracy to which the inspiratory: expiratory ratio meter could be read did not make this possible.

**Circuit D.**

The results of the previous investigations indicated to us that the combination of a similar expiratory pathway to that of circuit C with an improved method of delivery of fresh gas to the lungs could be the basis of an improved circuit.

The circuit shown in figure 8 was developed. The fresh gas inflow from the anaesthetic machine is diverted, preferentially, into a concertina bag by a lightly loaded diversion valve. The concertina bag of 200 ml capacity is enclosed within a bottle of 1,500 ml capacity and is compressed by the inflation pulses of a Newcastle ventilator. The driving ventilator was also modified to sense the cycling pressure at a point on the expiratory limb of the T-piece. This converts the system into a solely pressure-cycled device and prevents any biphasic inflow pattern and prevents the development of excessive airway pressures that are theoretically possible with circuit B. The driving ventilator will cycle when the pressure in the airway has reached a predetermined level and the compressed gas in the bottle will then pass to atmosphere via the Fink valve (Fink, 1954) which is included on the inflation line from the ventilator. This valve ensures early closure of the exit
pathway from the bottle during inflation and rapid opening as the ventilator cycles so that the bottle pressure falls and the bellows are free to refill. A degree of occlusion of the inflating line between the valve and the point of insertion of the tube operating the valve does not interfere with the function of the valve but enables flow into the bottle during the inflation phase to be slowed to a variable degree. This permits control of the duration of the inspiratory phase over a wide range.

The fresh gas inflow line includes a non-return valve and a spring-loaded pressure-relief valve. Fresh gas flow is only diverted into the bellows during the expiratory phase, since during the inflation phase the non-return valve closes and the continuing fresh gas flow vents to atmosphere through the pressure-relief valve. The pressure-relief valve is lightly loaded so that when the bellows become fully distended, before the next inflation, the continuing fresh gas flow prefers to escape through this valve rather than via the diversion valve to the expiratory limb.

A two-way tap and bypass line are also fitted to the inflation line so that fresh gas may be diverted directly to the T-piece and the whole inflating assembly is bypassed. If an open-ended rebreathing bag is then attached to the expiratory limb the system is converted for manual or spontaneous ventilation. During manual compression of the rebreathing bag the diversion valve acts as a non-return valve and prevents gas entering the concertina bellows.

Compression of the concertina bag by the driving ventilator forces fresh gas past the diversion valve. The lungs are inflated since the T-piece is occluded by the power-operated valve within the driving ventilator. This valve operates synchronously with the onset of the inflation phase of the ventilator. It opens the expiratory pathway freely to atmosphere at the same time as the ventilator cycles with the completion of the inflation pulse. It is at this point that the bottle pressure falls steeply and the bellows are free to fill with fresh gas.

Figure 9 illustrates the characteristic flow and pressure relationships found with this circuit. The
rapid rise in inspiratory flow rate to peak value is coincident with the attainment of peak airway and bottle pressure. After the ventilator cycles at the end of phase 2 the bottle pressure falls rapidly, allowing the bellows to refill, and peak expiratory flow is attained without significant delay. The expiratory pathway is separate and opens widely to atmosphere. There is no dependence on falling bottle pressure or opening of a weighted valve. The airway pressure is able to fall rapidly and the expiratory flow is complete before the start of the next inspiration and there is no residual airway or lung pressure.

If the inflation line is occluded to a variable degree, the duration of inspiration can be lengthened allowing the necessary flexibility of control in the face of varying conditions of airway resistance and lung compliance.

The circuit ventilated the model lungs satisfactorily and in clinical use was easy to operate (table II). No delicate balance of controls is necessary. A conveniently measured fresh gas flow rate is chosen which must be in excess of the expected minute volume. The cycling pressure control is then adjusted to cause the delivery of a satisfactory tidal volume and the inspiratory time control is adjusted to suit the clinical circumstances.

**DISCUSSION**

**The inspiratory phase.**

In these studies no attempt has been made to produce an inspiratory flow wave form of a particular character. The wave forms resulting when the attachments are applied to model lungs have been recorded in an analysis of function. The ability to alter the duration of the inflation phase independently is felt to be of value since this enables the clinician to choose a type of inspiration which will result in the most effective ventilation. With system D it is possible to prolong the inspiratory time and this results in a trend towards a square wave inspiratory flow pattern and a slower rise in airway pressure (fig. 10). This capability for independent control over inspiratory time is only possible if the driving ventilator is sensing its cycling pressure on the expiratory line.

It is possible that some of the delay in cycling with circuit B between the start of the fall in bottle pressure and the start of expiratory flow from the lungs would be explained if there is delay in cycling of the Newcastle ventilator. A further series of studies, described in the Appendix, was carried out and this established that with circuit B cessation of the inflation phase and opening of the expiratory valve were synchronous. This cycling point occurs 0.02 sec after peak bottle pressure is achieved (fig. 11).

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**FIG. 10**

Prolongation of inspiratory time resulting from closure of screw clip in circuit D.

**FIG. 11**

Cycling time relationships, circuit B driven by Newcastle ventilator.

The biphasic character of the inspiratory flow pattern which occurs with system B causes a sustained peak lung pressure. Although it can be agreed that this will reduce the effective dead-space (Cooper, personal communication, 1966; Watson, 1962) such a wave form is undesirable in cases with any cardiovascular instability since it will reduce cardiac output. It is also to be noted that very little increase in tidal volume occurs whilst the lung pressure is maintained. The
biphasic character of inspiratory flow can be avoided in the volume-cycled device, (i) if the size of the enclosing bottle is not unnecessarily large, for then there will be little or no delay in cycling of the driving ventilator after the reservoir bag has emptied, and (ii) if the peak bottle pressure is adjusted to be only just greater than the peak airway pressure. Any tendency to delay in the opening of the expiratory valve will be lessened.

The expiratory flow character.

The rapid attainment of peak expiratory flow rate as soon as inspiration is complete can only be attained if there is a wide expiratory pathway which opens to atmosphere. This is impossible in circuits A and B because expiration is delayed by the falling bottle pressure, and in addition in system B, by a weight-loaded valve. The satisfactory character of the expiratory pathway in circuit D is due to a separate design similar in outflow character to a T-piece occluder (circuit C). The desirability of producing a subatmospheric phase during expiration is open to debate. We believe that if the outflow pathway is of good design there is no need to assist expiration and that it is undesirable that negative pressures should exist after the completion of expiratory flow. Although this may be an advantage to the cardiovascular system, negative lung pressures probably have undesirable pulmonary effects. It is difficult to arrange negative pressure which assists expiration but does not continue during the true expiratory:inspiratory pause.

The use of negative pressure generated by the driving ventilator as is possible with the Pulmoflator, in association with system B, must be considered. This will tend to overcome the delay in opening of the expiratory valve after the ventilator has cycled, since the bottle pressure will fall more rapidly and the airway pressure will soon exceed the bottle pressure. If this negative pressure is maintained, it will overcome the weight of the expiratory valve and residual pressures will not occur. It is important to realize that a driving ventilator with this facility is desirable when using system B and adds further to the complexity of the adjustments that have to be made. If too high a level of negative pressure is inadvertently used it is, theoretically, possible that this pressure could be transmitted to the diversion valve and lift it. This will result in failure of the bag to refill before the next inflation pulse and the operator must be aware that this risk exists.

An expiratory valve of good design incorporated in a separate outflow pathway overcomes these problems. The valve which forms an integral part of the Newcastle ventilator is operated from the same power source as the inflating pulse and so operates synchronously. The resistance to outflow is very low and equals 0.4 cm H₂O at 10 litres/minute.

The inspiratory:expiratory ratio.

Mushin, Mapleson and Lunn (1962) have deduced criteria for ventilating these small subjects and an I:E ratio of 1:2 is agreed to be desirable in clinical practice. Experimentally, in order to study events to completion during the expiratory phase a more extreme I:E ratio was chosen. The inevitable rise in airway pressure which results (Mushin, Mapleson and Lunn, 1962) was accepted. In practice, with systems A, C and D the inspiratory time can be adjusted to ensure I:E ratios of 1:2.

The fresh gas inflow.

Some of the disadvantages of system A can be minimized if the lowest possible fresh gas flow is used but in practice we tend to use a higher fresh gas flow than necessary so as to be certain that rebreathing will not occur. Harrison (1964b) has shown that in order to prevent rebreathing in these circumstances as high a fresh gas flow as 2.8 times the minute volume may be required. A lower fresh gas flow rate will still prevent rebreathing if there is a distinct expiratory:inspiratory pause, a slow end to expiration, or a slow start to inspiration. Low inspired carbon dioxide concentrations will, in this system, depend on a high fresh gas flow rate and the slow end to expiratory flow. These are features which we wish to avoid if expiration is to be through a low resistance pathway. A high fresh gas flow is also necessary with circuit C in order to provide suitable peak inspiratory flow rates.

Circuits B and D act as fresh gas flow collectors; it is intended that only fresh gas collected in this way be inflated into the lungs. Low fresh gas flows without wastage are therefore possible
and with circuit B there is considerable theoretical advantage in the ability to measure the minute volume from the fresh gas flow measured at the Rotameters. At the low flow rates required difficulties in accurate measurement and the accurate performance of a vaporizer arise. In the design of circuit D we have accepted some gas wastage and the sacrifice of the ability to measure minute volume in favour of accurately mixed gases and accurate vapour concentrations.

The inadvertent or misguided use of fresh gas flows in excess of the minute volume with circuit B will cause pressure cycling. If it is intended that volume cycling should occur and a high cycling pressure is set on the driving ventilator, these high gas flows will cause peak airway pressures to approach the bottle pressure and there will be added resistance to expiration as the excess gas “spills over” into the expiratory limb.

**Conversion to manual ventilation.**

An important feature of any ventilator circuit is that conversion to manual inflation or spontaneous respiration should be easily accomplished. It is desirable that the manual system should be a direct one enabling the operator to sense lung compliance. Direct attachment of an open-ended bag to the expiratory limb of the T-piece can be easily arranged with circuits A, C and D. In the latter case the bypass line is opened to convert the circuit to a conventional “Rees modification”. There is no way of easily converting circuit B unless an indirect means, using manual generation of the inflation pulses to the bottle, is used.

**APPENDIX**

**The time relationships involved during cycling of the Newcastle ventilator.**

The cycling mechanism of the Newcastle ventilator (Burn, 1967) is actuated by transmission of pressure sensed in either the inflating or the expiratory line to the control bag beneath the hinged flap. Once this pressure exceeds the combined effect of the weight of the flap and the variable attraction of a magnet to an armature on the under surface of the flap, the hinged flap rises. This action separates a second magnet from the valve housing on the 60 Lb./sq.in. line causing the slug within it to fall. This cuts off both the inflation pulse and the pressure sustaining the expiratory valve. Mechanically, the ventilator has cycled at this point.

Figure 11 demonstrates the time relationships involved. A valve was constructed which, when inserted into the 60 Lb./sq.in. line, would generate a signal indicating that the slug within the inflation valve housing had fallen and cycling had occurred. A pneumotachograph tracing indicates events on the expiratory line from the bottle of circuit B to the driving ventilator just proximal to the expiratory valve and the changes in the bottle pressure when the ventilator is attached to circuit B are shown.

The fall of the slug on the inflation line and the commencement of flow through the expiratory valve are shown to be coincident. Peak expiratory flow through the valve occurs 0.04 sec later. Peak bottle pressure occurs 0.01 sec prior to mechanical cycling and the significant fall in bottle pressure commences just before peak expiratory flow is reached. A flow of gas is shown to occur along the expiratory line, falling to zero flow just prior to opening of the expiratory valve, and is due to gas flowing into the control bag. When the Newcastle ventilator is used in conjunction with system D, the inflation side of the circuit is separated from the expiratory and “sensing” side of the circuit by the infant attachment. This introduces factors which alter the time relationships in the cycling sequence.

Under these circumstances, if a pneumotachogram is obtained from the side arm of the injector on the inflation line of the Newcastle ventilator, entrainment causes flow through it during the inflation phase (trace 1, fig. 12). Entrained flow decreases when the bottle pressure rises to a peak because of decreased injector efficiency in the face of higher pressure. Only when there is cessation of flow through the Venturi side arm can the ventilator be said to have cycled from a functional point of view. If there is any flow in this direction there must be gas passing through the jet of the injector and therefore still passing into the bottle of the attachment. Mechanical cycling, as indicated by the slug falling occurs 0.03 sec before this functional cycling point and the expiratory valve opens (trace 2, fig. 12) 0.02 sec after functional cycling. Flow out of the side arm of the entrainment port occurs because gas in the tubing proximal to the Fink valve exhausts
along this route to atmosphere. Sensing of the cycling pressure is taken from the expiratory line and as a result the changes in bottle pressure do not necessarily relate to cycling. As shown in figure 12, peak bottle pressure may be achieved 0.05 sec prior to mechanical cycling.

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REFERENCES
