A PRESSURE-CYCLED VENTILATOR WITH MULTIPLE FUNCTIONAL BEHAVIOUR

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SUMMARY

A pressure-cycled flow generator type of ventilator is described which can be adjusted to give a wide range of inspiratory flow patterns and therefore the inspiratory pressure wave so produced can also be altered. The expiratory flow pattern can be adjusted by means of an expiratory resistance. The laboratory and clinical performance of the machine and its nebulizers are described and suggestions made for the clinical use of the different modes of action of the ventilator.

The pressure-cycled constant flow generator type of ventilator which is powered by compressed gas can have many advantages, namely small size, a minimum number of components, an efficient nebulizer humidifier, and therefore a low total cost. Unfortunately, this type of ventilator is normally difficult to use at its maximum efficiency because the setting of the controls is complex and the functional behaviour of the ventilator after adjusting it is not always predictable, particularly because the controls usually have some interaction. Furthermore, the pressure and flow inspiratory wave forms tend to be either fixed or adjustable over too narrow a range. The British Oxygen Company produced a series of two-stage injectors (Mapleson, 1962) which give almost constant flow characteristics over the whole range of flows and resulting airway pressures (against which the injector must work) likely to be found in clinical practice. Figure 1, shows that at their worst the injectors may decrease the delivered flow by 24 per cent, but this will only occur at peak airway pressures so that over the major part of the inspiratory cycle during normal clinical use the injectors behave as almost constant flow devices. These injectors were developed for the Cyclator and it seemed that the operating principle of that machine could, by major re-design and development, become the basis of an efficient ventilator for the management of respiratory failure. If a ventilator is to be used in an intensive therapy unit it will be operated by nursing staff untrained in pulmonary mechanics; this means that although the initial settings will be made by the medical staff the minute-to-minute management must be left in the hands of the unit nursing staff. It is in this respect that a series of three injectors for flow control would seem to be an advantage because the choice of the correct flow range would be made by the medical staff and subsequent adjustment is likely to be of a minor nature and simple to perform with little effect on other settings.

These criteria of low cost (the eventual cost will be approximately £150), operational simplicity, and a wide control of functional behaviour were borne in mind in designing the ventilator which has the following mechanical principles.

DESCRIPTION

The new ventilator is essentially a pressure-cycled device incorporating a range of interchangeable inspiratory flow injectors which are constant flow generators. The reader is referred to figure 2 for the following detailed description of its principle of operation.

The ventilator is entirely pneumatic, being operated by air or oxygen at a pressure of 60 lb/sq.in. The driving gas passes through a pressure regulating device which functions as the inspiratory flow control before entering a chamber containing a magnetically activated rocker valve.

Flow performance of the three types of injectors against back pressure. Measurements made with Rotameters and a water manometer. The solid line indicates the range of flows obtained with each type of injector with the flow control set to a maximum driving pressure of 60 Lb./sq.in. The broken lines show the injector performance with the flow control set to a driving pressure of 35 Lb./sq.in. measurement under constant flow conditions.

Fig. 2
Schematic diagram of the machine function.
The exit from this chamber leads to the inspiratory phase injector, the nebulizer and to the expiratory valve. During inspiration oxygen passing through the injector will entrain atmospheric air through the air inlet filter and supplementary oxygen if this is required. The combined gases then pass a check valve, the function of which will be described later, and a non-return valve where it mixes with oxygen (or air) and atomized water or medicament from the nebulizer, before passing through the delivery hose to the patient. During the inhalation phase the gas pressure holds the expiratory valve closed. As pressure builds up on the patient side of the injector, it is also transmitted back to an expandable bellows causing it to move against the resistance of a spring which is adjustable by means of the inspiratory pressure control. Inside the bellows are two magnets, one fixed and one movable, and as the bellows expands the relative positions of these two magnets will change. The rocker valve, being made of ferrous material, is influenced by these magnets and will tip one way or the other and start or stop the gas flow according to which way the bellows moves. Thus it will be seen that during inspiration pressure will build up inside the bellows until it reaches a predetermined value set by the inspiratory pressure control, at which point the magnetic valve will shut off the flow of gas to the injector, the nebulizer and the expiratory valve. The small non-return valve inside the bellows prevents the rapid escape of gas and this now seeps out slowly through a needle valve which is the expiratory time control. Clearly as the bellows collapses the magnets will revert to their original position and the rocker valve will once again trip the mechanism into inspiration. It will also be apparent from the diagram that when the supply of gas is cut off at the termination of the inspiratory phase, the pressure in the expiratory valve balloon will decay rapidly due to its small volume, the gas leaking away through the entrainment orifice of the injector and the expiratory valve will open, permitting the patient's expiration to pass to atmosphere.

During expiration the ventilator acts as a constant pressure generator (atmospheric) due to the series resistance imposed by the expiratory valve (Mapleson, 1959). This resistance is manually adjustable in order to provide a means of retarding expiration. The expiratory phase is time-cycled and is dependent upon the escape of gas from the bellows past the needle valve of the expiratory time control.

A patient trigger is provided which can override the normal timing of the ventilator and trip the machine into inspiration. Any inspiratory effort by the patient will deflect a diaphragm in the trigger unit away from its seat against which it is held by a magnet. This permits gas from the bellows to be rapidly dumped to atmosphere regardless of the setting of the expiratory time control.

The patient trigger magnet is adjustable and permits the sensitivity of the trigger to be varied between about 0.5 and 3 cm H₂O. The check valve referred to earlier is in fact a double valve and its function is to impose sufficient resistance to the inflow of gas through the entrainment port to enable the full inspiratory effort of the patient to be applied to the trigger diaphragm. At the same time it permits reverse flow which is necessary when the contents of the bellows are being dumped to atmosphere.

The co-axial breathing hose assembly has been specially developed for use with this type of ventilator and has been described in detail previously (Bushman and Robinson, 1968).

It is considered necessary to maintain a constant entrainment ratio and therefore a constant inspired Po₂ (Mayrhofer and Steinbereithner, 1967; Robinson, 1967) and it is impossible to vary the rate of flow of an injector over too wide a range without altering the entrainment ratio. It is for this reason that three interchangeable injectors each with an entrainment ratio of 3:1 are used, giving adjustable flows of 10–15, 30–50 and 50–70 l./min for each injector when discharging freely to atmosphere. The constancy of the entrainment ratio for each injector was checked with a fast gas analyzer (Atlas Mass Spectrometer M3) by measuring the inspired oxygen concentration delivered by the ventilator to a conscious subject triggering the machine. It was shown that the three injectors produced over the entire range of flow and pressure settings available a constant inspired oxygen concentration of 40±3 per cent. However, with the nebulizer in circuit, during any single inspiratory phase there is an initial puff of a 5 per cent higher oxygen concentration.
lasting 100 m.sec. Similar investigations with the mass spectrometer have shown that it is possible to obtain a 100 per cent inspired oxygen concentration if additional oxygen is supplied to the entrainment port at a rate in excess of the injector flow. Concentrations below 100 per cent can be obtained with lower added oxygen flows in which case a constant inspired oxygen concentration is not obtained because the first part of the inspiratory phase has a 20 per cent higher oxygen concentration.

Two nebulizers have been developed for use with the ventilator; the first of these, a low output unit is based upon a design, the characteristics of which have previously been described (Wright, 1958). This unit is designed to inject nebulized water and drugs into the main gas stream via a side limb (fig. 2) and its low performance is a safety factor if drugs such as isoprenaline are to be given. The second nebulizer is an in-line unit with a considerably greater output and a correspondingly larger reservoir (200 ml) (fig. 7).

The performance of the two nebulizers was assessed by sampling gas delivered by the ventilator into a 1-metre length of 2.5 cm i.d. hose maintained at room temperature (24°C). Measurements of water in the vapour phase and the liquid phase (as micro droplets) being made with a mass spectrometer. The Wright type of nebulizer was shown to produce just over 50 per cent saturation of the gas at 24°C. However, when the gas was sampled from a respiratory hose maintained at body temperature (38°C) even the in-line nebulizer did not produce gas saturated with water vapour nor did any other commercially available gas-driven nebulizer tested in the same manner. The implications of these observations and the experimental methods employed are to be the subject of another communication.

The performance of the ventilator was assessed in two ways: first by ventilating a 5-gallon drum via a 20 cm long by 10 mm i.d. cuffed endotracheal tube. Such an arrangement has a compliance of 0.04 l./cm H$_2$O and an airway resistance of 2 cm H$_2$O at 0.5 l./sec. Airway resistance was increased to 12 cm H$_2$O at 0.5 l./sec by inserting a 7 cm length of 7 mm i.d. tubing into the endotracheal tube. The clinical assessment was made in normal adults after surgery, the recordings being made immediately before reversal of curarization.

In each series of experiments the nebulizer was in circuit and the airway pressure and that within the artificial lung were measured with CEC strain gauges and a Devices d.c. amplifier. Oesophageal pressure was recorded with a latex balloon of 0.06 mm thickness, 10 cm long, with a circumference of 3.5 cm and inflated with 2.0 ml of air. Flow was sensed with a Fleisch pneumotachograph head coupled to a Statham differential strain gauge and flow was integrated with time to volume via a Devices integrator. Recordings were made with a Devices hot-wire recorder and pressure/volume and flow/volume loops recorded on an X-Y plotter.

**RESULTS**

The performance of the machine against the test rig described with a 20 mm long tube i.d. 10 cm showed that the design specifications were upheld and, as expected, the machine was found to be very dependent on the flow resistance against which it was working. In this context it should be noted that resistance to gas flow is partly dependent upon the rate of flow, so that for the same resistance low flows should produce a decrease in the pressure difference across the resistance. Bearing in mind that the machine cycles at the airway pressure, if the test rig is used with a high airway resistance, then at the moment of cycling, the pressure in the lung should equal the cycling pressure provided low inspiratory flows are delivered. These conditions are exemplified by the traces A and B in figure 3. In each case the machine was set to the same cycling pressure and expiratory time. The two tracings show a slight difference of 2 cm H$_2$O between the airway pressures due to pressure drop along the inspiratory hose.

Trace A, obtained with the high flow injector in which flow was set to maximum by use of the flow control, shows 15 cm H$_2$O difference between the cycling pressure and that within the lung. The tidal volume can be seen to be 750 ml.

Trace B shows the effect of using the same high flow injector with a low setting of the flow control to overcome the airway resistance. The lung pressure now more nearly equals the cycling pressure with a consequent increase in the tidal volume to 1100 ml.

Results obtained during the patient study con-
Flow, airway and lung pressures and tidal volume obtained with the high flow injector at high and low flow settings on the artificial lung.
FLOW, AIRWAY AND OESOPHAGEAL Pressures and Tidal Volume Obtained in a Patient Study Using the High and Low Flow Injectors.
firmed these laboratory findings. Figure 4, trace C shows the effect of the high flow injector, with flow control set to maximum. It can be seen that the flow decreases a little up to the cycling pressure of 42 cm H₂O and that there is a steep rise in airway pressure until airway resistance is overcome at 25 cm H₂O, when the predicted flow into the lungs commences and the oesophageal pressure begins to rise. Trace D obtained with a low flow injector shows that this low flow overcomes airway resistance at a lower pressure of 10 cm H₂O and that inspiratory time is lengthened to 3.6 sec. It should be noted that expiratory time remains constant and that although the cycling pressure remains constant the low flow with longer inspiratory period has resulted in an increase in tidal volume of 650 ml. The low flow has increased the mean intra-oesophageal pressure from 6 to 9 cm H₂O and the end expiratory oesophageal pressure no longer reaches atmospheric (4 cm H₂O) which probably results from the increase in end expiratory lung volume occurring when low inspiratory flow rates are used (Bushman and Robinson, 1968). Furthermore, examination of the lung compliance loops obtained at the same time showed an increase in the compliance slope from 0.19 to 0.23 l./cm H₂O when the low flow injector was used.

An adjustable patient triggering device and an efficient nebulizer are useful features which have previously been discussed (Robinson, 1967).

**DISCUSSION**

Constant flow generators which are pressure-cycled can be difficult to use because airway resistance may cause the airway pressure to rise precipitously and cycle the ventilator before an adequate quantity of gas has entered the lungs. From the work presented it is concluded that these defects can be overcome by the ventilator described because of the wide range of flows available from the three constant flow injectors which permit the inspiratory time and flow to be adjusted to provide the desired tidal volume. In clinical practice this can be readily achieved by placing a Wright respirometer at the expiratory end of the hose in front of the expiratory valve and adjusting the inspiratory flow and pressure controls until the appropriate tidal volume is reached. A change in airway resistance or compliance during use will be evident by a decrease in length of the inspiratory phase resulting in an increase in the respiratory rate.

Constant flow generators which are pressure-cycled tend to have a fixed pressure and flow pattern but by use of a control to decrease the flow of gas to the injector this machine can produce a constant flow or a flow pattern which decreases as inspiration continues. It can therefore produce either a steeply rising or a plateau form of inspiratory pressure.

The inclusion of an expiratory resistance in the design of this machine requires some explanation. It might be thought that decreasing expiratory flow by the addition of an expiratory resistance would increase the end expiratory lung volume (FRC) because the end expiratory oesophageal pressure would also be raised. However, the resistance chosen (10 cm H₂O at 0.5 l/sec) is insufficient to have this effect provided sufficient expiratory time is allowed. This can be seen from scrutiny of the traces in figure 5. Trace E, with the resistance, shows no change in end expiratory lung volume occurring when low inspiratory flow rates are used (Bushman and Robinson, 1968). Furthermore, examination of the lung compliance loops obtained at the same time showed an increase in the compliance slope from 0.19 to 0.23 l./cm H₂O when the low flow injector was used.

An adjustable patient triggering device and an efficient nebulizer are useful features which have previously been discussed (Robinson, 1967).
Flow, airway and oesophageal pressures and tidal volume obtained with (E) and without (F) the expiratory resistance during a patient study.

**Fig. 5**
being ventilated who have generalized obstructive lung disease.

The ability to exercise further control over the flow by decreasing the efficiency of the injector in such a way as to cause a 25–40 per cent drop in flow during the latter third of inspiration was deliberately embodied in the machine. It was realized that if flow could be decreased to very low values in the latter part of inspiration then the machine could be almost stalling before the cycling pressure was reached, thus giving an almost flat latter portion of the inspiratory pressure curve. It is known (Torres, Lyons and Emerson, 1960) that low flow rates with an airway pressure plateau allow better distribution of inspired gas by preventing overinflation of areas of lung which have high time-constants. This can be achieved by the flow pattern produced by the low flow injector with a decreased driving flow. Furthermore, if viscous forces in the thorax have to be overcome, as, for example, in patients with critically crushed chests, pneumatic consolidation, etc., then with an inspiratory flow pattern which falls in the latter third of inspiration this plateau form of the inspiratory pressure pattern will be more effective in dissipating such forces because these have been shown to be time-dependent (Robinson, 1967). Furthermore, it was demonstrated that this type of flow pattern produced a higher end expiratory oesophageal pressure which can only be due to an increase in the end expiratory lung volume. Therefore this type of flow pattern moves the subject's lung up to a steeper portion of the lung compliance curve so that an increase in volume for the same airway cycling pressure is obtained. This was confirmed by the change in the slope of the lung compliance loop. In this respect there is some evidence that there is a fall in compliance during pressure ventilation (Laver et al., 1964) and this may be associated with an increase in the A-a gradient for oxygen. This may be prevented by intermittent overinflation or by using large tidal volume (Bendixen and Laver, 1965). Possibly, by increasing compliance in the manner described, the A-a Po_2 difference may be decreased although the authors have been unable to demonstrate this in the series of patients studied.

Be that as it may, the ability to provide large tidal volumes is useful in the treatment of patients with crushed chest and pneumatic consolidation or collapse. However, while such large volumes result in a fall in arterial PCO_2 to an extent that might be deleterious, this is easily prevented with the ventilator described. If large tidal volumes are required then the inner hose of the twin axial hose may be shortened to give an appropriate amount of deadspace. Up to 300 ml can be easily obtained and it is a simple matter to prepare hoses with a calibrated deadspace.

CONCLUSION

An attempt has been made to design a pressure-cycled constant flow generator type of ventilator having a low cost and with simple controls which would have an adaptability of flow and pressure patterns not usually found with this type of machine. It is considered that most of the aims have been met. No attempt has been made to
provide a subatmospheric expiratory phase as this would add to both complexity and cost. Furthermore, the use of subatmospheric phase is only required during severe cardiovascular depression and this must be treated directly so that pressure ventilation no longer affects cardiac output. Also two of the authors (J.S.R. and J.B.; work in progress) have evidence suggesting that distribution of inspired gas is adversely affected by the use of a subatmospheric phase during expiration.

Success with artificial ventilation of the lungs can often be ascertained by the reduction in the A-a Po$_2$ difference. The constant entrainment characteristics make this easier as the inspired concentration of oxygen can be reliably estimated. In addition, the facility to produce 100 per cent inspired oxygen concentration can be used to remove the apparent alveolar-arterial gradient which is due to gas distribution errors (Berggren, 1942) and so to assess the contribution to the A-a gradient for oxygen due solely to intrapulmonary shunting.

The machine will shortly be commercially available and has the general appearance seen in figure 7.

REFERENCES


