

## The Physical Properties of Ventilators in the Inspiratory Phase

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In order to examine the mechanical characteristics of the inspiratory circuit of respirator, the latter is assumed to be equivalent to a gas pressure source ( $P_s$ ), having an internal resistance ( $R$ ) and an internal compliance ( $C$ ). A method for determining  $P_s$ ,  $R$  and  $C$  for a given respirator when the pressure source is either constant or sinusoidal is described. The procedure consists essentially of determining the relationship between the rate of flow delivered by the apparatus and the outlet pressure while the respirator is loaded by an external, gradually-varying resistance. Results obtained with five instruments, showing striking differences among them, are presented. These measurements can provide data for theoretical studies as well as for clinical use of the apparatus.

A VENTILATOR can be studied two ways: by measurement of its physical properties, or by determination of its clinical efficiency as shown by improvement in blood gases. Although the latter is obviously useful, the interpretation of results is difficult because of the diversity of patients and casual factors such as cooperation in awake subjects. Physical measurements of ventilators avoid these difficulties and can provide useful information concerning the ability of a given apparatus to assure proper artificial ventilation under a wide variety of pathologic conditions.

Controlled ventilation can be defined, on physical grounds, as periodic gas transfer between two mechanical systems, the ventilator and the thorax. The transfer is related quantitatively to the physical properties of both systems. Thus, to predict the efficiency of an apparatus for patients with given

mechanical disorders, it is of interest to know its physical characteristics. Furthermore, this knowledge allows one to compare ventilators, as well as to use actual physical data in theoretical or analog studies.

The work of Mushin *et al.*<sup>1</sup> contributed significantly to this field by defining qualitative mechanical characteristics of ventilators and providing functional analyses of them. Concentrating on the inspiratory phase, Mapleson<sup>2</sup> described two extremes of ventilators. In the first, a "flow generator," the temporal flow pattern during the inspiratory phase is determined entirely by the apparatus, whatever the thorax-lung properties of the patient connected to it. In the second a "pressure generator," the ventilator determines only the pattern of imposed pressure, while the rate of flow can vary widely according to the mechanical characteristics of the respiratory system. Because minute ventilation is related to, among other factors, the inspiratory flow rate, it is obvious that a given change in the physical features of the lung, for instance, an increase in airway resistance, will not affect ventilation in the same way with both types of apparatus.

Such a general classification of respirators according to mechanical properties is, therefore, useful, and can even be used for theoretical studies.<sup>3</sup> However, it provides only qualitative, schematic information regarding the ventilators. In fact, a number of instruments cannot be classed as either purely flow or purely pressure generators. It is of interest, therefore, to define mechanical characteristics that are easy to assess quantitatively and are suitable for a large variety of instruments. Those described in this paper concern the inspiratory circuit of ventilators, but can also be used to some extent for the expiratory

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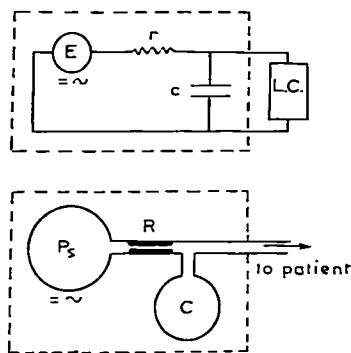


FIG. 1. *Top*: Schematic representation of an electric power supply consisting of a generator of electromotive force ( $E$ ) with an internal resistance ( $r$ ) and an internal capacitance ( $c$ ).  $L.C.$  is the external load circuit. *Bottom*: Mechanical analog of the electric power supply. A respirator is assumed to be equivalent to a gas pressure source ( $P_s$ ), having an internal resistance ( $R$ ) and an internal compliance ( $C$ ).

side. Cycling mechanisms for change-over from inspiration to expiration and *vice versa* are not considered.

### Choice and Definition of Mechanical Characteristics

Considering only the inspiratory phase, there is a close analogy between a ventilator and an electrical power supply. The first is intended to provide gas flow to a system which has certain mechanical properties (resistance, compliance and inertia), the second to provide current to a system which has electrical properties (resistance, capacitance, and inductance). As in the case of a ventilator, the output voltage and current—analogs of pressure and rate of flow—delivered by an electrical supply depend upon both the properties of the receiving system (load circuit) and its own characteristics. It is of interest, therefore, to extend the comparison and to assume an analogy between the characteristics of both systems.

As shown in figure 1 (top) a power supply can be represented schematically by a generator of electromotive force ( $E$ ) with an internal resistance ( $r$ ), shunted by a capaci-

tance ( $c$ ). For a given value of  $E$  and a given load circuit, current and voltage at the output are related to  $r$  and  $c$ . Keeping in mind the analogy between electrical and mechanical quantities (table 1), a ventilator can be assumed to be equivalent to a gas pressure source ( $P_s$ ) loaded by a mechanical resistance ( $R$ ), and connected to the patient in parallel with a distensible system having a compliance ( $C$ )\* (fig. 1, bottom). The pressure source can be constant (steady pressure during the inspiratory phase) or variable, sinusoidal for example.

There is strong evidence that the mechanical properties of a number of ventilators can be described adequately in this way. In all cases, indeed, internal and external tubing connecting the gas supply to the patient imposes resistance to gas flow. Moreover devices such as needle valves, one-way valves, constrictions, etc., constitute additional, and sometimes high, resistances to gas flow.

Similarly, every ventilator has a more or less significant compliance, due, in particular, to compressibility of the gas in the inspiratory circuit: the greater the internal volume of the circuit (including tubing, chambers, storage bags, etc.), the greater will be the compliance. Elasticity of certain parts of the circuit, particularly diaphragms, walls of bellows, or bags, must also be considered. In some cases the compliance is very small ( $<1$  ml/cm water) but in others it is not negligible in comparison with the patient's lung-thorax elasticity and thus can affect the inspiratory pattern of flow considerably.

The power source is perhaps more difficult to consider as always being rigorously a

\* For simplicity, internal inductances of the power supplies, as inertial factors for the ventilators, very small in practice, are neglected.

TABLE 1. Analogies between Electrical and Mechanical Quantities

Electrical	Mechanical
Charge (Q)	Volume (V)
Current (I)	Flow ( $\dot{V}$ )
Voltage (u)	Pressure (P)
Electromotive force (E)	" "
Resistance (r)	Resistance (R)
Capacitance (c)	Compliance (C)

pressure source. In a number of instruments the source is a tank of compressed gas, a compressor, an injector or a concertina bag loaded by a weight or spring, all of which are (in general, constant) pressure sources in close approximation. However, in some cases the gas flow is delivered by a piston driven through a cylinder or by a concertina bag alternately compressed and expanded by a powerful electric motor. These devices can be thought to be essentially flow generators. In fact, such systems are mechanically equivalent to very high (constant or not) pressure sources loaded by high resistance.

Such considerations regarding the characteristics of ventilators provide evidence that, in most cases, as far as their mechanical properties are concerned, these instruments can be schematically described by the equivalent system proposed above. Further evidence for this will be provided by the results of measurements of several commercial ventilators.

### Measurement of Mechanical Characteristics

It is not possible to describe a complete procedure that allows one to determine in every case the physical characteristics defined above. Each ventilator is a complicated mechanical device. Careful study of the technical brochure and working diagram will always be useful to ascertain the most convenient method in each case. In order to obtain the nominal features of the apparatus, it is essential to eliminate leaks, for example, from safety valves. Generally, however, measurement of mechanical properties is easy, requiring only standard laboratory equipment such as pressure and flow transducers.

The first step is to determine if the pressure source is constant or variable. This, in general, is obvious from examination of the functional diagram, but easy to verify experimentally by recording inspiratory flow through a pneumotachograph with the outlet of the apparatus open to ambient air. If the pressure source is constant, gas flow reaches a steady level after a short rise-time and remains constant during the inspiratory phase. In contrast, gas flow must vary with time if the source is nonconstant, as sinusoidal.

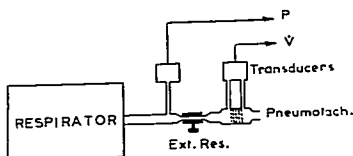


FIG. 2. Schematic representation of the system used to establish the flow-pressure diagram. A mechanical resistance is placed at the outlet of the inspiratory circuit and gradually varied while the pressure at the inlet of the resistance ( $P$ ) and the rate of flow ( $\dot{V}$ ) are measured simultaneously.

This first step is particularly important since the procedures to determine the driving pressure ( $P_d$ ), the internal resistance ( $R$ ) and the internal compliance ( $C$ ) differ according to whether the pressure source is constant or not.

### CONSTANT PRESSURE SOURCE

*Driving Pressure and Internal Resistance.* When the pressure source is constant, the determination of  $P_d$  and  $R$  can be done easily by studying for the inspiratory phase the steady-state relationship between gas flow delivered by the apparatus and the pressure at its outlet. For this purpose a mechanical resistance is placed at the outlet and varied gradually. For each value of this load, the rate of flow ( $\dot{V}$ ) and the pressure at the inlet of the resistance (loading pressure) are measured simultaneously (fig. 2). The values obtained for a wide range of loading pressures are then plotted on a flow-pressure diagram; measurements can be repeated for a variety of respirator adjustments.

To prevent the occurrence of expiration before a steady level of pressure and flow rate is obtained, it is necessary to set the duration of the inspiratory phase, the cycling pressure, or the cycling volume (according to the change-over mechanism) at a sufficiently high value.

A typical record obtained with the Bird Mark 7,\* with the corresponding flow-pressure diagram, is shown in figure 3. For a wide range of loading pressures, the relationship between flow and pressure is nearly linear; we found similar curves for all respirators tested when the pressure sources were constant.

\* Bird Corporation, Palm Springs, California.

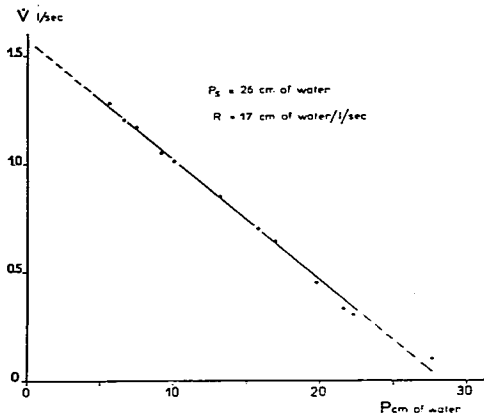
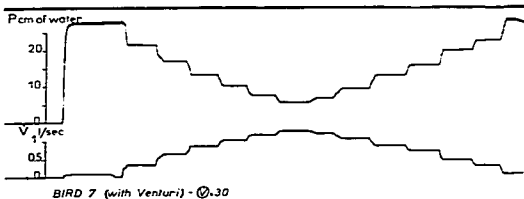


FIG. 3. Top: Record of pressure and inspiratory rate of flow obtained with the Bird Mark 7 when the external resistance was varied gradually. The flow control of the instrument has been set on 30 (arbitrary graduation). Bottom: Flow-pressure curve obtained from the record shown above. The relationship is nearly linear and rate of flow varies greatly with loading pressure.

To understand how such a curve allows one to determine  $P_s$  and  $R$ , one must consider the behavior of the model represented in figure 1. It can be shown (Appendix 1) that in static conditions, that is, when the system has reached equilibrium at a given external resistance, the steady rate of flow ( $\dot{V}$ ) observed is related only to the driving pressure and to internal resistance, being independent of compliance. Assuming  $R$  (the internal resistance) and  $r$  (the external resistance) to be viscous resistances, we have the relation:

$$P_s = (r + R)\dot{V} \quad (1)$$

or, with the loading pressure  $P = r\dot{V}$ :

$$P_s = R\dot{V} + P \quad (2)$$

and one obtains:

$$\dot{V} = \frac{P_s}{R} - \frac{P}{R} \quad (3)$$

Thus, as actually observed with ventilators with the simple mechanical analog there is a linear relation between rate of flow and loading pressure. This close agreement shows that the analog is valid and convenient. It indicates, in particular, that the internal resistance of the apparatus can be considered viscous in most cases.

From equation 3 we can see that internal resistance ( $R$ ) corresponds to the reciprocal of the slope of the flow-pressure curve ( $R = \Delta P_s / \Delta \dot{V}$ ), and thus is easily determined from experimental measurements. In like manner, equation 3 shows that the rate of flow becomes zero when  $P = P_s$ . The driving pressure is given by the observed loading pressure when the external resistance is sufficiently high to annul gas flow. This corresponds to the intercept between the flow-pressure curve and the abscissa. The driving pressure can

also be obtained from the intercept of the curve on the ordinate, since for  $P = 0$ ,  $\dot{V} = P_e/R$ . Thus,  $P_e$  is given by the product of the internal resistance and the observed flow rate when the load is zero.

For the example shown in figure 3, the internal resistance can be estimated to be about 17 cm water/l/sec and the driving pressure 26 cm water.

**Internal Compliance.** There are several ways to analyze the elastic properties of a constant-source respirator. First, when gas compressibility is thought to represent the major part of this elasticity, it is possible to obtain a rough estimate of the internal compliance by determining the approximate volume of gas contained in the inspiratory circuit. This can be obtained by measuring the diameters and lengths of internal and external tubing, the cross-section of bellows, etc. If  $V$  is the volume in liters, internal compliance is given by  $C \approx V/1,000$  l/cm water.

With the apparatus off, and no leaks in the inspiratory circuit, one can, with a syringe and a manometer, determine directly the compliance of the system by the volume of gas which must be introduced into the inspiratory circuit to obtain a pressure rise of 1 cm water.

In some cases, however, these procedures cannot be used. Consideration of the simple mechanical analog shown in figure 1 provides another method. Assuming that, at the

beginning of inspiration, the driving pressure rises abruptly from zero to a given value  $P_e$ , the rate of flow through an external resistance ( $r$ ) does not reach a steady level immediately but increases with time according to an exponential given in Appendix 1. The time constant  $\gamma$  of this phenomenon is given by

$$\gamma = \frac{rRC}{r + R} \quad (4)$$

After a time equal to  $\gamma$ , the rate of flow reaches 63.2 per cent of the terminal value and it reaches 86.5 per cent and 95 per cent of the equilibrium plateau for times equal to 2 and 3  $\gamma$ , respectively. The loading pressure increases, according to a similar law, with the same time constant.

It is then possible to determine  $\gamma$  by recording at high speed the variations in loading pressure or flow rate at the beginning of the inspiratory phase. Such a record is shown in figure 4. There is good agreement between the observed curves and theoretical variations of pressure and flow as given by the equation of Appendix 1. A mean value of  $\gamma$  is obtained on the pressure record from the time necessary to obtain the above percentages of the steady pressure level. Knowing  $r$  and  $R$ , one can easily compute  $C$ . This procedure provides a useful result if the response time of the measuring device is sufficiently short and if there are not too many artifacts on the records. It is preferable to have a rather high external

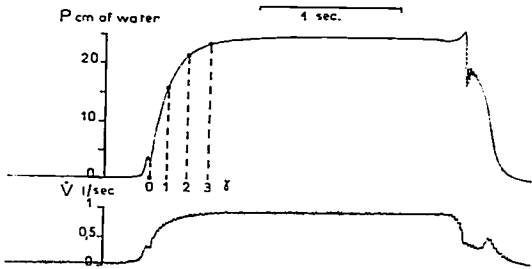


FIG. 4. Variation with time of the loading pressure and rate of flow during inspiration with the Bird Mark 7, a mechanical resistance having been placed at the outlet of the apparatus (external resistance  $r = 26.5$  cm water/l/sec). The time-constant ( $\gamma$ ) of the pressure rise is determined as described in the text.

BIRD MARK 7 -

Ext. Res ( $r$ ) = 26.5 cm of water/l/sec

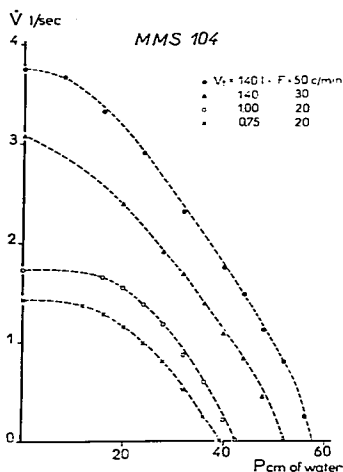


FIG. 5. Flow-pressure curves obtained with the MMS 104 (sine-wave generator) for different settings of the controls (tidal volume  $V_1$  and frequency  $F$ ). The observed relationships are curvilinear.

resistance ( $r$ ) in order to increase  $\gamma$  and to make its measurement easier.

#### NONCONSTANT PRESSURE SOURCE

In some respirators, in which the driving pressure varies with respect to time during the inspiratory phase, it can generally be considered to be approximately a sine-wave pressure source (Engström, MMS 104). We consider here the case where this approximation can be made.

As above, the mechanical characteristics of the system can be determined on a flow-pressure diagram, that is, by measuring simultaneously the rate of flow through an external, gradually varied, resistance, and pressure at the inlet. However, with such devices the rate of flow and thus the loading pressure, varies with the driving pressure according to a sine-wave pattern during inspiration. Thus, the system is never at static conditions and the flow rate through a given external resistance is related not only to  $P_s$  and  $R$ , but also to internal compliance  $C$  (Appendix

2). Thus, the relationship between the amplitudes of these two variables is more complicated.

As shown in Appendix 2, this relationship is linear only when the internal compliance is small and can be neglected. Hence, rate of flow is related to loading pressure, as indicated by equation 3. When such a linear flow-pressure curve is obtained with a sine-wave generator, compliance can be considered negligible. The amplitude of  $P_s$  and the value of  $R$  can then be computed according to the procedure given above.

More generally, the observed relationship is curvilinear, as can be predicted from the equation given in Appendix 2 when the internal compliance is significant. For example, figure 5 shows flow-pressure curves obtained with a sine-wave generator, the MMS 104,<sup>†</sup> for different adjustments of the apparatus.

If the internal compliance of the system cannot be determined separately, it can be computed from the flow-pressure curve accord-

<sup>†</sup> Matériel Médical et Sanitaire, 19 rue Mathis, Paris 19<sup>ème</sup>, France.

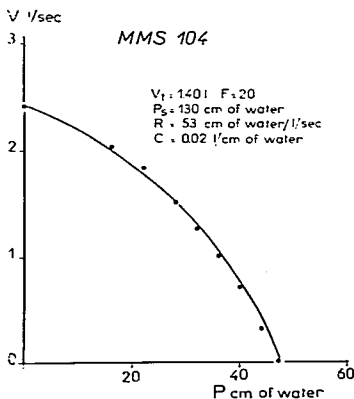


FIG. 6. Dots correspond to actual data obtained with the MMS 104 (sine-wave generator).  $P_s$ ,  $R$  and  $C$  have been computed from these data according to the equations given in the text. The solid line is the flow-pressure relationship computed for the simple mechanical model (fig. 1, bottom) using the same values of  $P_s$ ,  $R$  and  $C$ .

ing to the equation given in Appendix 2. Then,  $R$  and  $P_s$  can be derived from:

$$R = \frac{P_0}{(\dot{V}_0^2 - P_0^2 C^2 \omega^2)^{1/2}} \quad (5)$$

where  $P_0$  is the loading pressure necessary to give zero gas flow (intercept of the flow-pressure curve on the abscissa) and  $\dot{V}_0$  is the rate of flow observed when the loading pressure is zero (intercept of the curve on the ordinate). The angular velocity  $\omega$  of the sinusoidal phenomenon corresponds to the ratio  $\pi/I$ , where  $I$  is the duration of the inspiratory phase.

The amplitude ( $P_s$ ) of the driving pressure can then be obtained by:

$$P_s = \dot{V}_0 R \quad (6)$$

On the flow-pressure diagram shown in figure 6, the dots correspond to actual measurements obtained with a sine-wave ventilator. Values for the amplitude of the driving pressure,  $R$  and  $C$ , were computed from experimental results using the above equations. The solid line is the flow-pressure curve computed from the equation given in Appendix 2 for the mechanical model shown in figure 1. Good agreement exists between the theoretical and observed values, showing that the model provides a valid approximation of the apparatus.

#### MAXIMUM POWER

One can also derive from the flow-pressure curve the limiting (flow)  $\times$  (loading pressure) that can be achieved by the apparatus, that is, the maximum power ( $\dot{W}_m$ ) that can be dissipated by a viscous load. This parameter is related to  $P_s$  and  $R$  since, from equation 3:

$$\dot{W} = P \cdot \dot{V} = \frac{P(P_s - P)}{R} \quad (7)$$

It can be shown that  $\dot{W}$  reaches a maximum for  $P = P_s/2$ . Then:

$$\dot{W}_m = \frac{P_s^2}{4R} \quad (8)$$

The maximum power can be computed with this equation or obtained directly from the flow-pressure curve.

These equations are convenient for constant pressure devices, but not for sine-wave

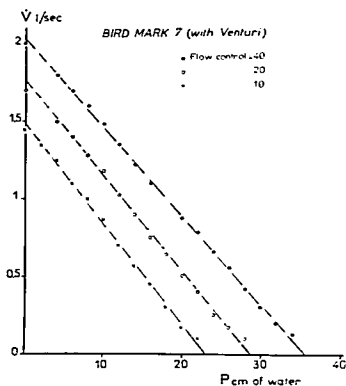


Fig. 7. Flow-pressure curves obtained with the Bird Mark 7 (Venturi in operation) for different settings of the flow control. All curves are nearly linear.

generators. With the latter type of apparatus, even with a viscous load, the rate of flow and the loading pressure vary with time. The mean value of the dissipated power is then given by the product of effective flow ( $\dot{V}/\sqrt{2}$ ) and effective pressure ( $P/\sqrt{2}$ ), that is, by  $P \cdot \dot{V}/2$ , where  $P$  and  $\dot{V}$  are the amplitudes of loading pressure and rate of flow, respectively. The maximum power can be derived by finding on the flow-pressure curve the maximum value of  $P \cdot \dot{V}$  and dividing the result by 2.

Since it is related to the two most important properties of the apparatus, this parameter provides a good idea of performance. However, with some respirators, the maximum power is reached only for very high loading pressures, sometimes above 100 cm water. Therefore, it is often more useful to determine on the flow-pressure curve the greatest product  $P \cdot \dot{V}$  that the apparatus can produce with a loading pressure equal to or less than a certain value, for example 40 cm water ( $\dot{W}_{40}$ ).

#### Results

According to the procedures defined mechanical characteristics of five commercial respirators have been determined. Four are

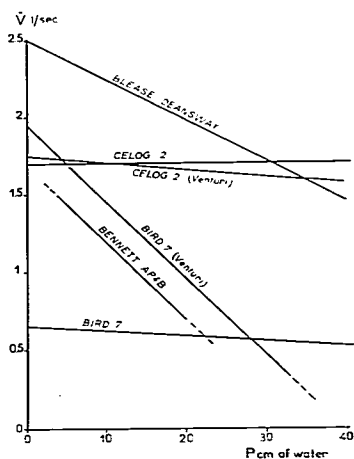


FIG. 8. Flow-pressure curves obtained with four respirators. Controls are adjusted to obtain the maximum rates of inspiratory gas flow.

constant pressure devices: the Bennett AP 4 B,\* the Bird Mark 7, the Blease Deansway,† and the Celog 2.‡ One is a sine-wave generator, the MMS 104. In each case, measurements have been made and characteristics computed for a wide variety of control adjustments.

Flow-pressure curves obtained with the MMS 104 appear in figure 5. In figure 7 a family of curves from the Bird Mark 7 is presented. With this apparatus the mechanical properties differ considerably according to whether the gas dilution system (Venturi) is used or not (fig. 8). The data shown in figure 7 were obtained with the Venturi in operation. As noted above, the observed curves are nearly linear, allowing one to determine  $P$ , and  $R$  easily. For different adjustments of the flow control, the driving pressures (given by the intercepts of the curves on the abscissa) vary greatly, while the internal resistance (reciprocal of the slope)

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† Anaesthetic Equipment Ltd, Deansway, Chessham, Bucks, England.

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is nearly constant. Varying the flow control, in fact, modifies the rate of flow at the inlet of the injector and, thus, the pressure source.

Flow-pressure curves obtained with the four constant pressure generators are shown in figure 8, the controls being adjusted to obtain the maximum inspiratory rate of flow. For the Bird Mark 7 and the Celog 2, results with and without the gas dilution systems are presented. The corresponding mechanical characteristics are given in table 2.

There are striking differences among the devices studied. Two the Bennett AP 4 B and the Bird Mark 7 (with Venturi) are equivalent to a low-pressure source with a small internal resistance. Variations of gas flow as a function of loading pressure, reciprocal of  $R$ , are very large: the apparatuses correspond to the pressure generators described by Mushin *et al.*,<sup>1</sup> since the rates of flow can vary widely according to the mechanical characteristics of the load system. Finally, the maximum power developed for a loading pressure equal to or less than 40 cm water is small.

In contrast, the Bird Mark 7 (without Venturi) and the Celog 2 (with and without Venturi) are equivalent to high- or very-high-pressure sources with large internal resistances. Variations of gas flow with external load are very small, and these respirators correspond to the flow generators defined by Mushin; the rate of flow is nearly independent of the loading system. The maximum power is high for the Celog 2, while small with the Bird because the driving pressure is not very high with respect to internal resistance.

For the Blease Deansway and MMS 104 instruments, the pressure sources are in the middle range, but internal resistances are not very high. Hence, variations of gas flow with loading pressure are relatively important, and these ventilators can be considered pressure generators. For the MMS 104 the internal compliance is not negligible with respect to lung-thorax system compliance and can affect the temporal pattern of gas flow during artificial ventilation.

## Discussion

It is useful to re-emphasize the exact meaning of the parameters defined above.



TABLE 2. Mechanical Properties of Respirators with Controls Adjusted to Obtain the Maximum Rate of Flow\*

Respirator	Equivalent Driving Pressure (cm water)	Equivalent Resistance (cm water/l/sec)	Equivalent Compliance (l/cm water)	Max. Power ( $P \leq 40$ ) (kg·m/sec)
Bennett AP 4B	35	21	—	0.14
Bird 7 (without Venturi)	175	250	0.002-0.003	0.21
Bird 7 (with Venturi)	40	18	0.002-0.003	0.19
Blease Deansway	100	38	—	0.58
Celog 2 (without venturi)	>1000	>1000	<0.001	0.68
Celog 2 (with venturi)	300	180	<0.001	0.62
MMS 104	85 (120)	32	0.011	0.38 (0.75)

\* For the MMS 104, sine-wave generator, effective values are given; peak values are indicated in brackets.

Rather than being real mechanical features of the respirator, they are those of an equivalent schematic system that behaves approximately the same way under particular circumstances, namely on a viscous load. In fact, the relationships between flow rate and loading pressure can be slightly different for the respirator and its analog when the load is a complex one, as is the case of the thoracopulmonary system. Moreover, for some respirators the actual physical properties can be thought to vary somewhat during the inspiratory phase. With a sine-wave genera-

tor, for instance, the gas volume contained in the inspiratory circuit varies with time, as does compliance. In this case the value obtained from the flow-pressure curve corresponds approximately to the mean compliance of the system.

I believe that measurement of mechanical characteristics as defined in this paper provides useful information regarding the performance of respirators. First, it can help the anaesthetist to choose the most convenient instrument for a particular circumstance. For instance, it is obvious that a respirator

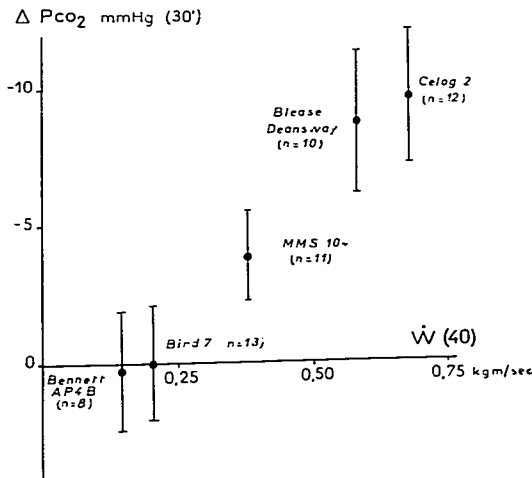


FIG. 9. Comparison of the maximum powers of five respirators and the improvement of  $PCO_2$  obtained in patients with chronic bronchitis after 30 minutes of artificial ventilation. Mean values of  $\Delta PCO_2$  plus and minus one standard error are indicated. The number of patients ventilated with each instrument is given in brackets.

which has low internal resistance and small power is unable to provide a large steady hyperventilation in patients whose resistances are high and variable, but the same apparatus may be well suited to ventilate the lungs of patients whose mechanical properties are nearly normal.

To verify these facts the maximum powers ( $\dot{V}_{40}$ ) of five respirators and the improvement in  $P_{aCO_2}$  obtained in patients with chronic bronchitis after 30 minutes of artificial ventilation have been compared. All patients had chronic hypercapnia (mean  $P_{aCO_2} = 52$  mm Hg,  $\sigma = 8.6$ ) and were tested after recovery from acute respiratory failure. Ventilation was administered by mask with 40 per cent  $O_2$  in air. As shown in figure 9, only the more powerful devices allowed notable reductions in  $P_{aCO_2}$  in patients who had high thoracopulmonary resistances.

On the other hand, the mechanical model given in figure 1 can be used for theoretical or analog studies of artificial ventilation. Knowing  $P_s$ ,  $R$  and  $C$ , it is possible to simulate in close approximation the main features of an apparatus during the inspiratory phase, and then to compare the behavior of different devices under given conditions.

This paper deals with mechanical properties of respirators for the inspiratory phase only. To a large extent, similar characteristics can be defined for the expiratory phase, the procedure being basically the same. Taking into account the cycling mechanisms for the changeover from inspiration to expiration, it is possible to determine completely the mechanical features of any respirator.

#### APPENDIX 1

Let us consider the simple mechanical model shown in figure 1, loaded by an external resistance  $r$ . At the beginning of inspiration the source pressure varies stepwise from zero to  $P_s$ . The rate of flow through  $r$  is given by the relation:

$$\dot{V} = \frac{P_s}{r + R} (1 - e^{-rR/(r+R)t}) \quad (9)$$

Thus, flow varies with time according to an exponential law. After equilibrium, that is, when  $t \rightarrow \infty$ :

$$\dot{V} = \frac{P_s}{r + R} \quad (10)$$

which is identical to equation 1.

#### APPENDIX 2

When the driving pressure varies with time according to a sine-wave pattern, the simple mechanical model of figure 1 gives the following relationship between the rate of flow ( $\dot{V}$ ), the properties of the ventilator ( $P_s$ ,  $R$  and  $C$ ) and the loading resistance ( $r$ ):

$$\dot{V} = \frac{P_s}{(r^2 + R^2 + 2rR + r^2R^2C^2\omega^2)^{1/2}} \quad (11)$$

In this equation  $\dot{V}$  and  $P_s$  represent, respectively, the amplitudes of the flow rate and driving pressure. Assuming a viscous external resistance, the amplitude of the loading pressure ( $P$ ) is equal to  $r\dot{V}$  and:

$$\dot{V} = \frac{-P + (P_s^2 - P^2 - P^2R^2C^2\omega^2)^{1/2}}{R} \quad (12)$$

This equation relates the rate of flow to the loading pressure. When  $C$  is very small and can be neglected, we obtain:

$$\dot{V} = \frac{P_s - P}{R} \quad (13)$$

which is the same as equation 3.

If the respirator actually behaves like the simple mechanical model, its compliance can be computed from the flow-pressure curve by the relation:

$$C = \frac{[(2\dot{V}_0\dot{V}_1P_0P_1)^2 - (\dot{V}_0^2P_0^2 - \dot{V}_1^2P_0^2 - \dot{V}_0^2P_1^2)^2]}{2\dot{V}_1P_1P_0^2\omega} \quad (14)$$

where  $\dot{V}_0$  is the flow rate observed when the loading pressure is zero (intercept of the flow-pressure curve on the ordinate),  $P_0$  the loading pressure necessary to give zero gas flow (intercept of the curve on the abscissae),  $\dot{V}_1$  and  $P_1$  the coordinates of any point taken on the curve.  $C$  is given in l/cm water if the flow rates are in l/sec, pressures in cm water and  $\omega$  in radian/sec.

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