THE MECHANICAL LUNG VENTILATOR AS ANOTHER "BLACK BOX"*

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IN OCTOBER, 1933, at the Twelfth Annual Congress of Anesthetists in Chicago, Dr. Harry J. Shields presented his paper "The Graphic Method of Teaching Anesthesia with Particular Emphasis on Respiration."¹ The substance of his paper is indicated by its title; namely, that the visual and audible appraisal of a patient's respirations enabled one to ascertain rather precisely the depth of ether anaesthesia. To illustrate his point graphically he showed two hand-drawn slides of the respiratory patterns of a patient struggling through Guedel's levels of ether anaesthesia (Figs. 1 and 2). Dr. Shields warned that "... new drugs and new anesthetic methods . . ." eventually would minimize the usefulness of these helpful respiratory signs in judging levels of anaesthesia. Today, 33 years later, his prophecy is confirmed. The use of new agents with pharmacologic signs much more subtle than those of ether, and a barrage of supporting drugs, are common in contemporary practice. In addition to Dr. Shields' accurate projection about new agents, the varying respiratory patterns he described are now "fixed" by the control knobs of the mechanical lung ventilator.

Last year Dr. John F. Nunn had the honour of presenting this lecture. I was intrigued by the title "The Lung as a Black Box."² As I read his introductory paragraph I realized that it clearly expressed my concept of a mechanical lung ventilator. With Dr. Nunn's kind permission (and that of the *Canadian Anaesthetists' Society Journal*) I borrowed part of his title and, with a few minor changes denoted by brackets, his first paragraph, which so ably serves as an introduction to my own topic:

"According to current usage, a 'black box' is a process or device whose internal workings are not understood by the operator, but whose function is important and may be studied in detail without appreciation of the internal mechanism. A typical example of a black box is the [mechanical lung ventilator] which is understood by very few of the population although its performance is important to the [patient]. Most of us dislike operating black boxes since it is intellectually unsatisfying and, on the practical level, places [the patient] in a position of great vulnerability if the mechanism should fail. However, at some period during the industrial revolution it became impossible for all men to understand the workings of everything they used and, from that date, we have become surrounded by an ever-increasing number of black boxes. Furthermore, we may confidently predict

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that the trend will increase as simple devices are replaced by their more complicated successors."

Although I have little understanding of the internal mechanical workings of the black boxes to be discussed, I hope to present a partial assessment of some of their functional capabilities.

CLASSIFICATION OF MECHANICAL LUNG VENTILATORS

The literature of the past decade relates numerous attempts to classify the various types of mechanical "black boxes." Wittenberger³ categorized all devices which use varying gas pressures to distend the lungs and thorax as *automatically cycled breathing machines*. These include not only those machines which produce automatic rhythmic pressure variations in the upper airways but also those which function by varying the pressure about the body—the iron lung, cuirass, and belt. We need not consider the latter since their usefulness has dwindled.

The automatically cycled breathing machines which are attached to the upper airway of the patient are called *lung ventilators*.⁴ These are assigned a variety of generic classifications depending upon the background and interest of the writer. Elam⁵ adopted the physiologic terms *volume-limited*, *pressure-variable* and *pressure-limited*, *volume-variable*, which relate the performance of the ventilator to changes in the lung-thorax compliance of the patient.

Fairley⁶ used a modification of Elam's terminology—*volume-constant*, *pressure-variable* and *pressure-constant*, *volume-variable*. The distinction between *limited* and *constant* needs semantic clarification.

Hunter's⁷ classification *volume-preset* and *pressure-preset* represents further simplification and focuses attention more directly on the ventilator function. He emphasizes however, that *volume-preset* denotes the ability of the device to compensate for diminishing compliance and increasing airway resistance while *pressure-preset* denotes the machine's ability to compensate for leaks.

Mapleson's⁸ classification relates to the cycling mechanism of the various ventilators. He divides the action of all machines into four parts; (1) an inspiratory phase, (2) change-over from this phase to the expiratory phase, (3) an expiratory phase, and (4) change-over from the expiratory phase back to the inspiratory phase. The change-over or cycling may be governed by time, pressure, or volume and may occur at the change from inspiration to expiration or from expiration to inspiration; hence, the classifications *time-cycled*, *pressure-cycled* and *volume-cycled*. The change-over from expiration to inspiration may be automatic or patient triggered. In addition, Mapleson separates all ventilators into two general types based on their behaviour during the inspiratory phase. When the ventilator determines the flow pattern (or volume) it is a *flow generator* and when it determines the pressure pattern it is a *pressure generator*.

FIGURES 1 and 2. Dr. Shields' graphic representation of the stages and levels of ether anaesthesia emphasizing the slow respiratory rate and expiratory pause characteristic of the second plane.

Engström⁹ states that classifying ventilators according to their cycling or triggering process is insufficient; further, he claims the Engström unit will not fit into any of the categories listed by Mapleson. Norlander¹⁰ used the terms *volume-controlled* and *pressure-controlled*. He believes that the *volume-controlled* ventilators should be classified according to "the driving force" applied to the ventilating gases. The terms *constant-force generator* with either *adjustable* or *non-adjustable flow* and *increasing-force generator* with either *direct* or *indirect* action are suggested. He neither tested nor classified the pressure-controlled apparatuses.

In an effort to make this discussion of "black boxes" as simple as possible Hunter's classification (*volume-preset* and *pressure-preset*) has been adopted. It can be seen that the performance of mechanical lung ventilators is usually assessed from simple pressure, flow, and volume patterns developed during clinical use; major emphasis is directed toward delivering the required tidal volume at the lowest possible mean mask pressure. Engström and Norlander¹¹ pointed out, however, that while the mechanical work accomplished by the respirator could be calculated from the pressure-volume diagram this does not reveal the momentary *power* output or capabilities of the ventilator. Knowledge of this aspect of respirator function is important since some pathological conditions of the lung require considerably more power for adequate delivery and distribution of the tidal volume within a stipulated time. Thus power curve plots are of considerable value in evaluating the mechanical capabilities of lung ventilators.

The following standard physiologic symbols are used throughout this presentation.

<i>Symbol</i>	<i>Term</i>	<i>Unit</i>
V	Volume of gas	L. or ml.
V _T	Tidal volume	L. or ml.
$\dot{V}$	Flow	L./min. or ml./sec.
$\dot{V}_A$	Minute alveolar ventilatory volume	L./min. or ml./min.
P	Pressure	cm. H ₂ O
$\bar{P}$	Mean pressure	cm. H ₂ O
F	Force	kg.
W	Work	kg.m.
Po	Power	kg.m./sec.
C	Compliance	L./cm. H ₂ O
f	Respiratory frequency	no. breaths/min.
i/e	Inspiratory/expiratory ratio	

In a simple system where the reacting force is constant, work is the product of the acting force and the distance over which it acts:

$$W = F \times D$$

Since power is the rate at which work is done, then

$$Po = W/t = (F \times D)/t.$$

FIGURE 3. Cartoonist's concept of a simple formula for work. In the left illustration the absent-minded iceboat helmsman has neglected to cast off his elastic mooring line; the work done is equal to the force of the wind times the distance the mooring line is extended within elastic limits. In the graph below, work is represented by the triangular area to the left of the dashed line. The right-hand illustration presents a similar situation with the addition of the work required to overcome the frictional resistance imposed by water (depicted by the hatched area below the dashed line on the accompanying graph). In both instances inertial resistance is neglected.

In a fluid pressure-volume system, work is equal to the product of pressure and flow so long as the conditions of pressure and flow remain constant:

$$W = P\dot{V}$$

and

$$P_o = W/t = P\dot{V}/t.$$

The resistive components in pulmonary ventilation are such that pressure-flow conditions do *not* remain constant; therefore, work and power must be expressed by the integral forms

$$W = k \int_{t_0}^{t_1} \Delta P \dot{V} dt, \quad P_o = \frac{k}{t} \int_{t_0}^{t_1} \Delta P \dot{V} dt$$

The term k is a constant relating to the units in which flow is expressed: 0.01 where flow is in L./sec. and .00017 where flow is in L./min.; ΔP is the change in intrapulmonary pressure in cm. H_2O ; $\dot{V}$ is flow in L./sec. or L./min.; t is the time in seconds or minutes; W is the work in kg.m.; and P_o is the power in kg.m./sec.

FIGURE 4. Cartoonist's concept of the integral forms for work and power in a fluid pressure-volume system. Hopefully the motorist possesses sufficient power to perform the work needed to inflate his tire. His pumping efforts must overcome elastic, frictional, and inertial resistances. These resistances are similar to those encountered by a mechanical lung ventilator; however, the inertia of the lung-thorax system is so small that it is usually neglected. Occasionally lung ventilators used in the operating room must overcome part of the weight of one or more tired surgeons; in this instance inertial resistance becomes meaningful.

In the integral forms, work is the *average* work done over an interval of time and power is the *instant* power at any moment. In respiratory physiology the work of breathing is usually expressed in kilogram metres and power in kilogram metres per second. A kilogram metre of work is equivalent to 9.806 joules or 9.806×10^7 ergs. One kilogram metre per second of power is equivalent to 9.806 watts or .013 horsepower.

METHOD

A "test lung" similar to that used by Elam,⁵ Saklad,¹² Fairley,¹³ and Norlander¹⁰ was constructed from two 20-litre bottles. The tubings, Y-connector, and stopcock leading from the pneumotachograph head to the bottles were of non-distensible material. The stopcock was arranged so that one bottle could be excluded from the system, thereby halving the compliance. Changes in pressure within the system were detected by a Statham P62D-250 transducer. Airflow was detected by a Statham Model 270 differential pressure transducer connected to the two sides of a pneumotachograph. The airway pressure strain gauge was calibrated

with a water manometer. The pneumotachograph was calibrated with a Fisher-Porter precision flowmeter.

Figure 5 is a block diagram of a data collection, processing, and recording system. With the exception of the special computer* and integrator,† the

FIGURE 5. Block diagram of the data sensing and processing system for recording pressure, flow, power, and volume or work. (When bottles are employed as test lungs, one side of the differential strain gauge is open to air.)

amplification and recording was done with a Sanborn Series 150 multichannel recorder. Tidal volumes were obtained by electronic integration of the flow rate signal. Instantaneous power was determined from the product of the pressure and flow signals using a "Hall effect" multiplier similar to that described by Engström and Norlander.¹¹ Work was obtained by time integration of the power signal.

The ventilators tested were those used on the respiratory ward and in the operating rooms. No claim is made as to their state of repair or functional perfection. Each ventilator was adjusted to deliver 0.5 L. to the 40-litre test lung. When possible the duration of inspiration was one second and the respiratory frequency was 20 per minute. Under these conditions pressure, flow, power,

*Constructed by A. L. Novak, Electrical Engineer, Department of Anesthesiology Research Laboratories, University of Minnesota Medical Center.

†Constructed by Medtronic, Inc., Minneapolis, Minnesota.

work, and volume were recorded; compliance of the test lung was then halved as the recording of the above data continued. Only the inspiratory phases of the ventilators were analysed. Minute "alveolar" ventilatory volumes were calculated by assuming an arbitrary dead space of 150 ml.

RESULTS

The waveform characteristics of the parameters studied appear in Figures 6a through 17a, pages 170 to 181; the full compliance traces appear to the left, and the half compliance to the right of the scale. The effects of halving compliance on T_v , W , $\bar{P}$, " $\dot{V}_A$," f , and i/e ratio are depicted on the bar graphs in Figures 6b through 17b.

Tidal volume. The ventilators are listed in order of decreasing ability to maintain V_T at half compliance. Of the volume-preset ventilators the Harvard Pump, Engström, Ohio Deluxe, Van Bergen, and Bennett BA 4 continued to deliver at least 90 per cent of the control V_T ; the Air Shields Ventimeter/Ventilator and Bird Mark VII-IV in the volume-preset mode delivered 80-90 per cent and the Air Shields Electronic 70 per cent. The pressure-preset ventilators (Bird Mark VIII without air entrainment, Bird Mark IX Servo, Bennett PR 2, and Bird Mark VIII-IV in the pressure-preset mode) delivered approximately 50 per cent of the control V_T at half compliance.

Inspiratory pressure contours. The ventilators arbitrarily are placed in two groups based on the magnitude of mean inspiratory pressure at full compliance: those developing mean pressures of less than 10 cm. of H_2O (favouring circulatory dynamics) and those developing mean pressures of more than 10 cm. H_2O (favouring lung gas distribution). The Van Bergen, Ohio Deluxe, Bird Mark IX Servo, Harvard Pump, Bird Mark VII-IV in the volume-preset mode, Air-Shields Ventimeter/Ventilator, Bennett BA 4, and Air Shields Electronic fall in the first group. The Bird Mark VIII without air entrainment, Bird Mark VIII-IV in the pressure-preset mode, Bennett PR 2, and Engström are in the second group.

Among those ventilators capable of delivering 90 per cent or more of the control V_T when compliance was halved, the Van Bergen and Ohio Deluxe develop relatively low mean pressures and the Harvard, Bennett BA 4, and Engström relatively high ones.

Flow curve. At the onset of inspiration, peak or near-peak flow is achieved almost instantaneously by all the ventilators except the Harvard Pump and the Van Bergen. The inspiratory flow curve of the Harvard Pump is similar to a half sine wave while that of the Van Bergen is a gradual incline to end-inspiration. The flow curve of the Engström is unique in that flow practically ceases after the first half of the inspiratory cycle.

Instant power plot. The Engström, Van Bergen, and Harvard Pump attained the highest instant power peaks at both full and half compliance. The instant power peak achieved by the Engström is higher than indicated because the galvanometer swung beyond its accuracy limit for the attenuation used.

Work. The average work performed by each ventilator during inspiration is indicated in the bar graphs. The mean average work required to introduce 500 ml. into the 40 L. volume was .043 kg.m. The small deviations from this mean reflect

the difficulties encountered in adjusting the inspiratory volume to exactly 500 ml. and the inspiratory time to precisely one second.

Respiratory frequency and i/e ratio. Considerable difficulty was experienced in adjusting the ventilators so that the duration of inspiration was one second and the respiratory frequency 20/minute. When this could not be accomplished inspiration was set as closely as possible to one second, with frequency and expiratory time deviating from the established scheme. The Bird Mark VIII without air entrainment requires 1.5 seconds to deliver 500 ml.; when air entrainment is used it is capable of delivering the required V_T in one second. Of the pressure-preset ventilators the Bennett PR 2 is unique in that it maintains a constant respiratory frequency and i/e ratio when compliance is halved.

DISCUSSION

The purpose of this investigation was to evaluate a series of "black boxes" used in our hospital. It was assumed that each machine was in functional order but no claim could be made that they were mechanically faultless. The experiment was designed to yield a comparative assessment of their performance. The elastic loads imposed by the test-lung at full and half compliance were comparable to those encountered in patients with lung disease.

The volume-preset ventilators failed to maintain the initial tidal volume at reduced compliance due to several factors: compression of gas within the "dead space" of the ventilator, increased internal leakage at higher pressures, and inability of the machine to develop sufficient power. The Engström, Ohio Deluxe, Van Bergen, Harvard Pump, and Bennett BA 4 maintained good volume constancy at half compliance. The pressure-preset ventilators could not be expected to do this.

Reduction in the calculated hypothetical " $\dot{V}_A$ " at half compliance is meaningful in judging the function of each machine; however, this factor can be used to compare one ventilator with another *only* when cycling frequencies and inspiratory/expiratory ratios are identical.

The calculated mean pressures were derived from the inspiratory portion of respiratory cycle. In evaluating the over-all effect of intermittent positive pressure breathing on cardiovascular and respiratory function the entire ventilatory cycle must be considered. The ventilators which produce low mean pressures disturb cardiovascular dynamics minimally but may not provide even alveolar gas distribution. Those producing high mean pressures favour alveolar distribution but may cause a fall in cardiac output. Since these "black boxes" frequently are used on patients with abnormal lungs and unstable cardiovascular systems, good judgment must be exercised in selecting appropriate pressure and flow rate waveforms. Frequently, the ill effects of uneven ventilation may be offset by increasing the tension of oxygen in the inspired gases.

Mechanical ventilators should be capable of adjusting their work load to the changing compliance of a patient's lungs. The power curves identify those ventilators which were capable of doubling their power output when compliance was halved. Machines which are incapable of this "servo" adjustment must be

monitored and adjusted frequently. This usually entails multiple arterial blood sampling for oxygen and carbon dioxide tensions. There are instances in which a patient's lungs are so "stiff" that a majority of available ventilators cannot provide the maximum power output required for adequate ventilation.

This report alone cannot serve as a standard for judging ventilators since only a fraction of the available ventilators were studied and these were subjected to only one test, namely, that of halving compliance. Comprehensive guidelines for the selection of "black boxes" may be found in the literature.^{14,15} However since one of the fundamental requirements of a suitable mechanical lung ventilator is that it delivers *safely* the *precise minute alveolar ventilatory volume* required to maintain normal arterial blood levels of oxygen and carbon dioxide with *minimal disturbance to cardiovascular function*,¹⁶ this study does identify in some units serious limitations which may curtail their ability to satisfy that requirement.

CONCLUSIONS

Twelve mechanical lung ventilators were evaluated from the standpoint of their ability to perform against a twofold increase in load. An analysis of the changes occurring in V_T , $\bar{P}$, $\dot{V}$, P_o , W , " $\dot{V}_A$," f , and i/e ratio revealed that only five of the volume-preset units could continue adequate ventilation under the increased load condition.

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RÉSUMÉ

La présente investigation concerne les divers ventilateurs employés dans notre hôpital. Nous avons considéré que chaque appareil à ventilation employé était en bon ordre, mais nous ne prétendons pas qu'ils aient été mécaniquement sans défauts. Le but de l'étude a été de comparer les modalités de leur rendement. Les résistances élastiques imposées par le poumon-échantillon à compliance complète ou partielle ressemblent aux résistances rencontrées chez les malades présentant une pathologie pulmonaire.

Les ventilateurs à volume prédéterminé ne purent maintenir le volume courant initial à compliance réduite à cause de plusieurs facteurs: compression des gaz à l'intérieur de l'espace mort du respirateur, augmentation des fuites aux pressions élevées, inefficacité de l'appareil à développer suffisamment de puissance. Les respirateurs Engström, Ohio Deluxe, Van Bergen, Harvard Pump, et Bennett BA 4 ont maintenu une ventilation de même volume à demi-compliance. Les ventilateurs à pression prédéterminée ne pouvaient évidemment accomplir ceci.

Une diminution dans le " $\dot{V}_A$ " hypothétique calculé à demi compliance est significatif pour juger de la fonction de chaque appareil; toutefois ce facteur ne

peut être employé pour comparer un ventilateur à un autre que lorsque les fréquences de cycle et les rapports inspiration-expiration sont identiques.

Les pressions moyennes ont été calculées à partir de la portion inspiratoire du cycle respiratoire. En évaluant l'effet global de la ventilation à pression positive intermittente sur les fonctions cardiovasculaire et respiratoire, le cycle ventilatoire entier doit être considéré.

Les ventilateurs à basse pression moyenne troubleront peu la dynamique cardiovasculaire, mais il peut arriver qu'ils ne procurent pas une distribution gazeuse alvéolaire homogène. Ceux qui produisent des pressions moyennes élevées favorisent une bonne distribution alvéolaire, mais ils peuvent diminuer le débit cardiaque. Comme ces appareils sont souvent employés chez des malades souffrant de pathologies pulmonaires et cardiovasculaires, il faut choisir avec discernement la pression et le volume appropriés.

Souvent, les inconvénients d'une ventilation inégale sont compensés par une augmentation de tension d' O_2 dans les gaz inspirés.

Les ventilateurs mécaniques devraient être capables d'ajuster leur puissance à la variation de compliance des poumons du malade. Les courbes de puissance font reconnaître les ventilateurs capables de doubler leur puissance lorsque la compliance diminue de moitié.

Les appareils incapables de s'ajuster "à demande" doivent être fréquemment contrôlés et ajustés. Ceci nécessite des analyses sanguines fréquentes pour établir les tensions d' O_2 et de CO_2 .

Certains patients présentent une telle résistance à la ventilation qu'un bon nombre parmi les respirateurs disponibles ne peuvent assurer la puissance nécessaire à une ventilation adéquate.

Cette étude seule ne peut servir de critère pour juger les respirateurs, puisque seulement quelques uns des respirateurs disponibles ont été comparés et que ceux qui ont été étudiés n'ont été soumis qu'à une seule épreuve, celle de la demi-compliance. La littérature peut aider à cette classification.^{14, 15}

Toutefois comme un des desiderata fondamentaux d'un bon ventilateur est d'assurer avec sécurité une ventilation alvéolaire minute capable de maintenir les paramètres sanguins en O_2 et CO_2 normaux sans nuire à la fonction circulatoire,¹⁶ cette étude indique chez certains appareils des limitations qui mettent en doute leur capacité de satisfaire à cette exigence.

FIGURE 6. *Harvard Pump (Animal Ventilator).* (a) The model tested has a fixed respiratory frequency. The pressure curve approximates a smooth sine wave at both compliances. The pressure peak is 18 cm. H_2O at full C and 34 cm. H_2O at half C. The V curve is also a smooth sine wave with a peak $\dot{V}$ of approximately 40 L./min. at both compliances. The maximum P_0 achieved at full C is .084 kg.m./sec. and at half C is 0.16 kg.m./sec. (b) This simple piston-cylinder pump maintains the initial V_T quite adequately at half C; $\bar{P}$ and W doubled. Values for f and the i/e ratio could not be varied on the model used but the duration of inspiration is consistent with the experimental plan. A fall of 10 per cent occurs in the " V_A ."

FIGURE 7. Engström Mod. 200. (a) At each compliance, inspiratory pressure increases rapidly to a plateau and remains there for 0.5 sec. The plateau is virtually flat at full C but inclines slightly at half C. The peak pressures are 17 and 32 cm. H₂O for full and half C respectively. The flow curve is unique in that near-maximum flow is reached almost instantaneously. At full C about 95 per cent of the volume is delivered in 0.4 sec., and at half C 95 per cent is delivered in 0.5 sec. In each instance very little $\dot{V}$ continued over the remainder of the one second inspiration. Like the $\dot{V}$ curve, the $\dot{P}$ curve peaks rapidly, achieving 0.184 kgm./sec. for full, and 0.264 + kgm./sec. for half C. (b) Initial V_r is fairly adequately maintained at half C; $\dot{P}$ and $\dot{W}$ nearly double. The i/e ratio could not be varied; this results in the inspiratory frequency of 18 when the inspiratory time is approximately one second. Calculated " $\dot{V}_A$ " falls approximately 6.5 per cent.

FIGURE 8. *Van Bergen Ventilator.* (a) The inspiratory pressure curve accelerates to a sharp peak and then falls rapidly. At full and half compliances the peak pressures are 16 and 30 cm. H_2O respectively. The $\dot{V}$ curve inclines steadily to a peak and then returns to zero instantaneously. At full C peak $\dot{V}$ is 48 L./min. and at half C, 42 L./min. The inclining $\dot{V}$ curve contains oscillations with frequencies of 16/sec. and amplitudes which increase from zero to peak flow. The inspiratory P_o curve accelerates to a peak of 0.12 kg.m./sec. at full C, and 0.20 kg.m./sec. at half C. (b) The initial V_T is adequately maintained at half C; P and W nearly double. The i/e ratio is constant, resulting in an f of 19 when i is one second. " $\dot{V}_A$ " falls 17 per cent.

FIGURE 9. *Ohio Deluxe Anesthesia Ventilator.* (a) The pressure curve has a sine wave configuration very similar to that of the Harvard Animal Respirator. At full C and half C the pressure peaks are 16 and 30 cm. H₂O respectively. The $\dot{V}$ pattern rises to peak instantly and then declines to zero in an irregularly oscillating manner. At full C maximum $\dot{V}$ is 46 L./min., and at half C 40 L./min. The P_o curve is jagged during its ascent and peak P_o is 0.072 kg.m./sec. at full, and 0.118 kg.m./sec. at half C. (b) V_t is well maintained at half C with W and $\dot{P}$ doubling; f drops to 19/min. with extension of the e by 0.1 second; " $\dot{V}_A$ " falls approximately 16 per cent.

FIGURE 10. Bennett BA 4 Anesthesia Assistor/Controller. (a) The inspiratory pressure pattern inclines convexly to peak and then falls sharply. Maximum pressures attained at full and half C are 17 and 34 cm. H₂O respectively. The $\dot{V}$ curve jumps to a notched peak, declines in linear fashion over most of the inspiratory phase, and then falls sharply to zero. The peak flows achieved are 38 and 32 L./min. for full and half C respectively. The inspiratory P_o curve is convex. At full C it peaks terminally at .076 kgm./sec. and at half C it peaks at .072 kgm./sec. pre-terminally and then declines slightly before terminating. (b) $\dot{V}_t$ is well maintained at the reduced C with W nearly doubling and $\dot{P}$ more than doubling. The change in the i/e ratio with an increase in i accounts for the high P at half C. W does not show a parallel increase to $\dot{P}$ since the $\dot{V}$ decreases. A 20 per cent drop in " $\dot{V}_A$ " occurs at half C.

AIR SHIELDS VENTILATOR /VENTILATOR C x I C x 0.5

FIGURE 11. Air Shields Ventilator/Ventilator. (a) The inspiratory pressure contour is a straight incline to peak and then falls rather sharply. At full and half C the pressure peaks are 17 and 29 cm. H_2O . The $\dot{V}$ pattern is a square wave with slight oscillations at its beginning. Peak flows are 30 and 29 L./min. for full and half C respectively. The P_o curve inclines in straight line fashion to a maximum power of .08 kg.m./sec. at full C and 0.12 kg.m./sec. at half C. (b) V_T falls 13 per cent at half C; P and $\dot{V}_A$ increase by 62 and 66 per cent respectively. Values for f and the i/e ratio remain constant. " $\dot{V}_A$ " falls approximately 20 per cent.

FIGURE 12. Bird Mark VII-IV A/C, volume-preset. (a) The inspiratory portion of the pressure curve inclines convexly and then plateaus briefly before falling off. Peak pressures are 16 and 26 cm. H₂O for full and half C. The $\dot{V}$ patterns reach their peaks in a jagged manner, then decline before terminating in an oscillating fashion. At full and half C the $\dot{V}$ peaks are 50 and 40 L./min. respectively. The P_0 curve inclines steeply, plateaus, and then falls sharply. Maximum powers of .064 and .096 kg.m./sec. are reached at full and half C. (b) With the Bird Mark VII set to deliver maximum $\dot{V}$ and Mark IV in the volume-preset mode the V_T falls about 16 per cent at half C. $\dot{P}$ doubles but W increases by only 66 per cent as a result of the decreased $\dot{V}$ at half C. Values for f and the i/e ratio do not change. " $\dot{V}_A$ " falls approximately 24 per cent.

FIGURE 13. Air Shields Electronic Ventilator. (a) The pressure pattern curves convexly to peak and then falls sharply. The maximum pressures attained at full and half C are 17 and 25 cm. H_2O . The $\dot{V}$ curve rises instantly to a peak, declines in straight line fashion, and then terminates sharply. The peak flows are 44 and 40 L./min. at full and half C. The P_o curve rises convexly to maximum powers of .068 and .06 kg.m./sec. for full and half C and then terminates abruptly. (b) $\dot{V}_T$ drops nearly 30 per cent at half C; there is very little increase in $\bar{W}$, with a proportionately greater increase in $\bar{P}$, indicating that $\dot{V}$ is reduced. Values for $\bar{P}$ and the $\dot{V}$ ratio remain unchanged. " $\dot{V}_A$ " falls 42 per cent.

FIGURE 14. Bird Mark VIII, without air entrainment. (a) The inspiratory pressure curve rises convexly to peak at 17 and 18 cm. H₂O for full and half C and then falls off rapidly. The V pattern is a square wave with the typical oscillations at the beginning and end of the plateau. A peak V of 26 L./min. is reached at both full and half C. The P₀ curve rises convexly, falls sharply, notches, and then tapers convexly to zero. Peak powers achieved at full and half C are .068 and .06 kg.m./sec. respectively. (b) While P remains constant, V_T, W, and "V_A" are approximately halved at half C. The i/e ratio and f are pressure dependent and vary with changes in C.

FIGURE 15. Bird Mark IX "Servo." (a) The configuration of the inspiratory pressure pattern is similar to that of the Bird Mark VIII. Peak pressures for full and half C are 17 cm. H_2O in both instances. The $\dot{V}$ curve with this unit is much smoother than with the Bird Mark VII or VIII and the oscillations are absent. $\dot{V}$ rises rapidly to peak, declines rather steeply, and then falls sharply to zero. Peak flows are 48 and 40 L./min. at full and at half C. The P_o curve is smooth and rounds nicely at peak before falling sharply. At full and half C the P_o peaks are .072 and .064 kg.m./sec. (b) With the $\bar{P}$ remaining constant the $\dot{V}_T$ falls nearly 50 per cent at half C. The average W performed during inspiration lessens markedly because the i shortens 33 per cent with f increasing from 17 to 20. The " $\dot{V}_A$ " decreases 66 per cent.

FIGURE 16. Bennett PR 2. (a) The inspiratory pressure wave is convex and approaches a square wave at the reduced C. Peak pressures reached at full and half C are 17 and 18 cm. H₂O respectively. The $\dot{V}$ pattern is virtually a sharp spike rising instantly to peak and then falling very sharply to become concave as it approaches zero flow. Peak flows attained are 72 and 68 L./min. at full and half C respectively. The duration of $\dot{V}$ is shortened when C is halved. The P_o curves have contours similar to the $\dot{V}$ patterns. Maximum P_o achieved at full and half C is .09 kg.m./sec. in each instance. (b) At half C, V_t and W are halved and $\bar{P}$ increases slightly. Values for f and the i/e ratio do not vary. " V_A " is reduced 70 per cent.

FIGURE 17. Bird Mark VIII-IV A/C in the pressure-preset mode. (a) The inspiratory pressure curve ascends convexly to peak and falls sharply. Peak pressures at full and half C are 17 cm. H₂O in each instance. The V pattern is characterized by an immediate sharp rise to peak and a gradual decline to zero flow. Oscillations occur near the end of the inspiratory phase. Peak flows achieved are 48 and 38 L./min. for full and half C respectively. Po curves are similar to those of the Bird Marx IX and maximum powers at both full and half C are .048 kgm./sec. (b) V_t and W fall 50 per cent when C is halved. A slight increase occurs in P. The value for f increases from 17 to 19, and i decreases. "V_d" is reduced about 70 per cent.

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