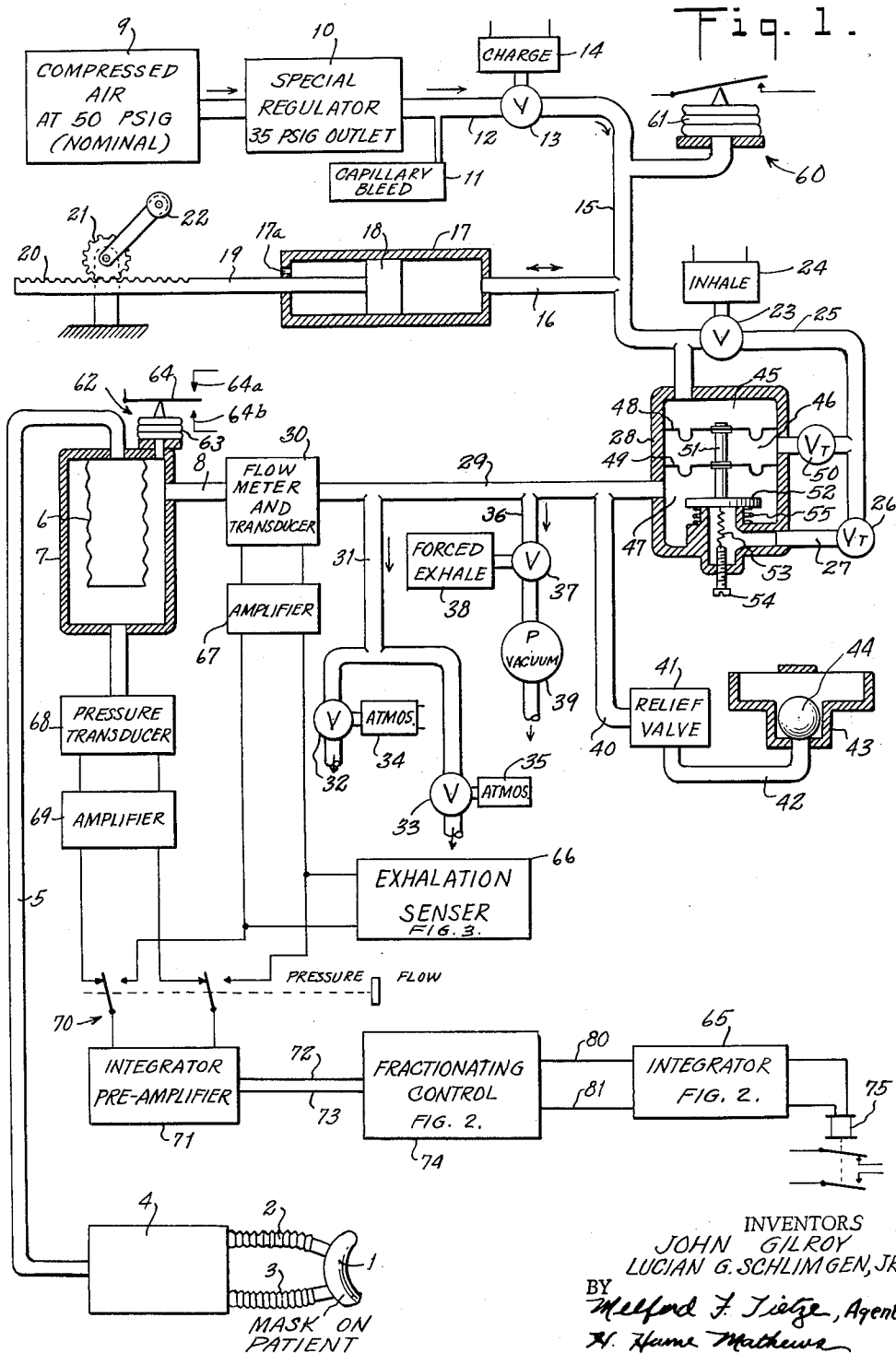


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5 Sheets-Sheet 1



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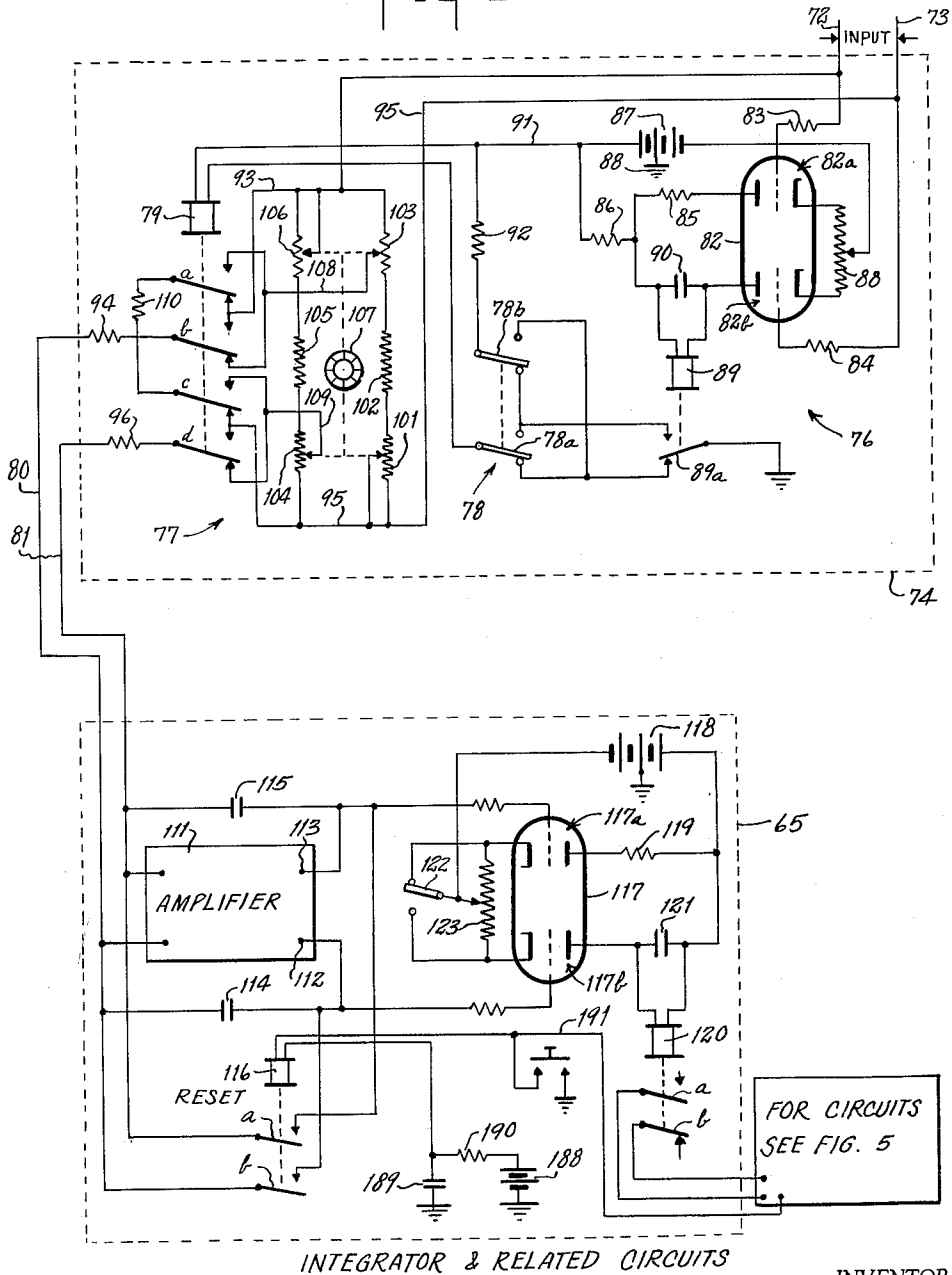
3,033,195

RESPIRATOR APPARATUS AND METHOD

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5 Sheets-Sheet 2

Fig. 2.



INTEGRATOR & RELATED CIRCUITS

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RESPIRATOR APPARATUS AND METHOD

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5 Sheets-Sheet 3

Fig. 3.

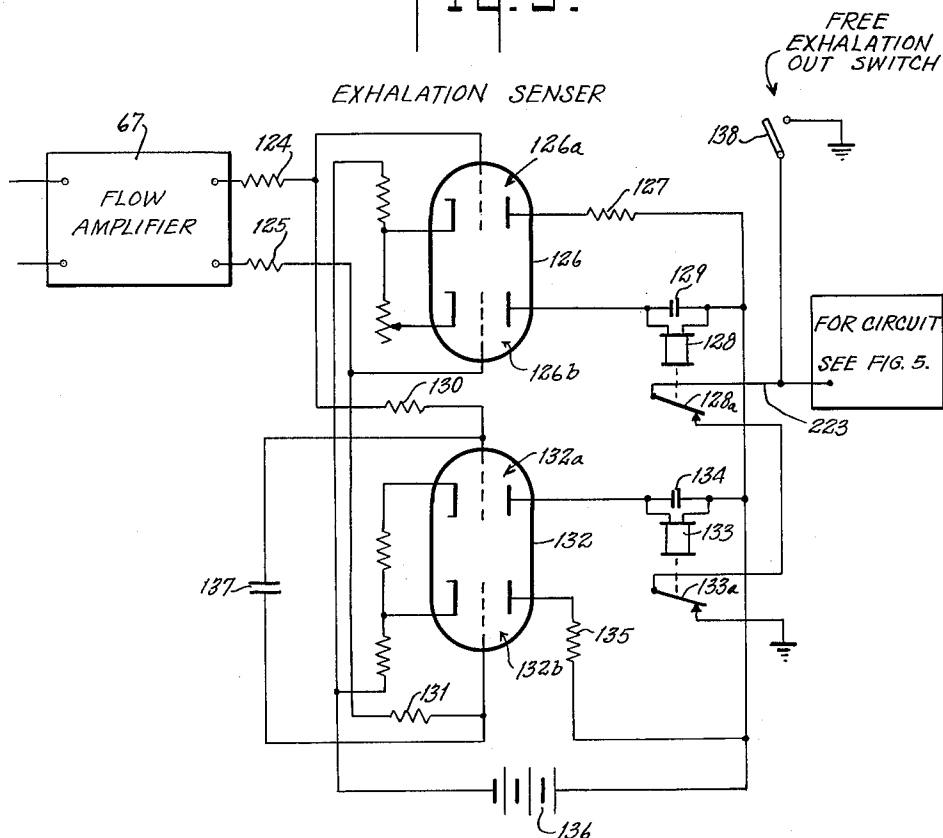
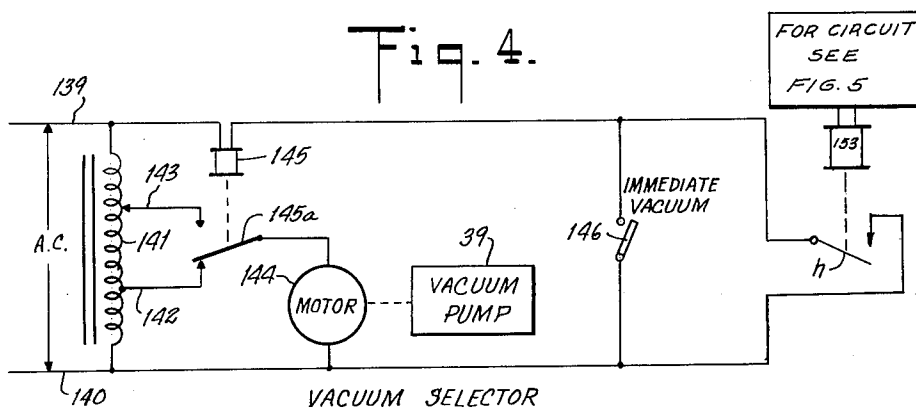


Fig. 4.



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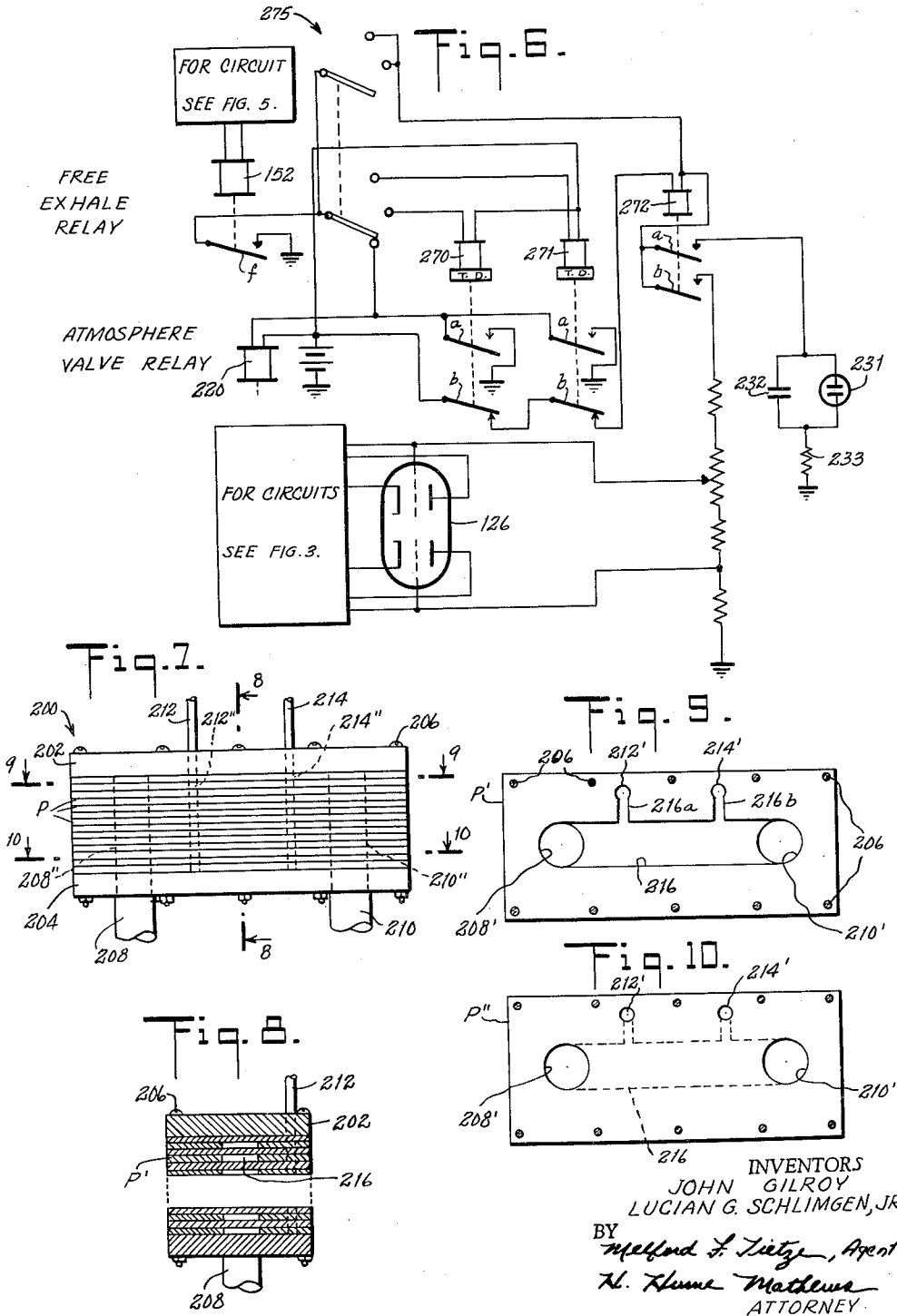
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5 Sheets-Sheet 5



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3,033,195

RESPIRATOR APPARATUS AND METHOD

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16 Claims. (Cl. 128—29)

This invention relates to respirator apparatus and particularly to apparatus which is settable or adjustable to accommodate a wide range of variations in different respiratory conditions.

Many different systems and apparatus for artificial respiration have been proposed and built. Most such systems have been concerned simply with accomplishing alternate inhalation and exhalation, without regard to the comfort of the patient or other subject of the respiratory apparatus and without regard to the patient's own efforts to breathe. The respirators of the prior art have not been readily adjustable to varying desired pressure or other conditions. Furthermore, they have not been adjustable to accommodate the differences between various patients, or even differences between the reactions of one patient at different times.

An object of the present invention is to provide an improved respirator apparatus.

Another object of the invention is to provide a respirator apparatus in which each of a number of variables may be independently adjusted, for the purpose of laboratory study of respiration or for the purpose of diagnosis and treatment of conditions in the respiratory tract of a patient.

Another object is to provide a respirator apparatus which is usable for anesthesia purposes.

Another object is to provide a respirator apparatus which supplies to the patient during inhalation a fixed volume of gas at a predetermined pressure or with variation over a predetermined range of pressures.

Another object is to provide a respirator apparatus which will supply the force necessary for inhalation and exhalation when the patient is making no effort to breathe, but which will follow and accommodate and supplement the patient's effort to breathe as soon as he makes any such effort.

Another object is to provide a respirator apparatus which will follow any of a number of breathing cycles, one of which may be selected by the operator of the apparatus.

Another object is to provide apparatus of the type described in which the breathing cycles are made up of a plurality of phases, including inhalation, exhalation and rest phases.

Another object is to provide respirator apparatus in which the phases of the operating cycle may be selected by the operator or determined by the patient's response.

Another object is to provide an improved respirator apparatus having provision for an exhalation phase and improved means for timing the duration of the exhalation phase.

Another object is to provide an improved respirator apparatus including improved means for sensing the direction of flow of gas to and from the patient.

Another object is to provide respirator apparatus including a means for sensing the true pressure in the patient's respiratory tract.

Another object is to provide improved respirator apparatus including improved control mechanism for the various phases and cycles involved. A further object is to provide such an improved control system including a relay for each phase, and a relay for each cycle, each cycle relay being effective to determine a particular sequence of phases.

2

Another object is to provide an improved respirator of the type described, in which the timing of the exhalation phase is determined by integrating the rate of flow of gas or the pressure of the gas supplied to or leaving the patient's respiratory tract.

A further object is to provide a respirator of the type described in which the various phases, cycles, durations of phases, and other variables are manually settable and controllable to produce variations in the respiratory program.

The foregoing and other objects of the invention are attained by providing a gas supply cylinder of fixed or adjustable volume, which is filled with air at a predetermined pressure before each inhalation phase. The air compressed in that cylinder is allowed to fall to a predetermined lower pressure, and the flow of air out of that cylinder is used to control the flow of respiratory gas to the patient. The flow of air to the cylinder is controlled by a charging valve. The flow of air from the cylinder is controlled by an inhalation valve. When the inhalation valve is closed, the patient may exhale either through a forced exhalation valve which supplies a negative pressure, i.e., vacuum to the patient or through an atmosphere valve which allows the patient to exhale freely.

The quantity of air inhaled by the patient is determined by the predetermined upper and lower pressures of the gas supply cylinder and by its volume. During inhalation, the flow or pressure to the patient is integrated with respect to time. After inhalation is completed, an exhalation phase is started, and a comparison integral of flow or pressure is taken until this integral reaches a predetermined ratio of the integral during inhalation. The integrator then terminates the exhalation phase and may establish either a rest phase or a new inhalation phase. Provision is made for controlling the predetermined ratio of the integrals.

The exhalation phase may be either free exhalation or forced exhalation, or it may be a free exhalation phase followed by a forced exhalation phase. The negative pressure or vacuum applied to the patient's respiratory tract may be controlled during the forced exhalation phase. Provision is made for sealing the patient's respiratory tract during a brief interval, for the purpose of measuring the alveolar pressure in the patient's lungs.

Other objects and advantages of the present invention will become apparent from a consideration of the following specification and claims, taken together with the accompanying drawings.

In the drawings:

FIG. 1 is a diagrammatic illustration of a respirator embodying the invention, including a block diagram of the principal electrical controls;

FIG. 2 is a wiring diagram of an integrator circuit and related circuits used in the electrical control system of the invention;

FIG. 3 is an electrical wiring diagram of an exhalation sensor circuit;

FIG. 4 is an electrical wiring diagram of a control circuit used in the apparatus of FIG. 1;

FIG. 5 is an electrical wiring diagram of the principal relay circuits controlling the respiratory cycles and phases of the apparatus;

FIG. 6 is an electrical wiring diagram of a circuit for producing a pressure measuring phase in the apparatus;

FIG. 7 illustrates a device, through which a gas stream whose flow is to be measured is adapted to be passed, having spaced pressure taps for deriving respective pressure signals whose differential is linearly related to the gas flow;

FIG. 8 is a vertical sectional view of the device shown in FIG. 10, taken along the line 8—8 and looking in the direction of the arrows;

FIG. 9 is a horizontal sectional view taken along line 9—9 in FIG. 7 looking in the direction of the arrows; and FIG. 10 is a horizontal sectional view taken along line 10—10 in FIG. 7, looking in the direction of the arrows.

GENERAL DESCRIPTION—FIG. 1

There is shown in FIG. 1 a respirator apparatus constructed in accordance with the invention. This apparatus includes a face mask 1 connected by flexible tubes 2 and 3 to a regenerated chamber 4, which is connected through a conduit 5 to a flexible walled chamber 6, commonly known as a breathing bag or rebreathing bag. The apparatus so far described is entirely conventional and allows the patient using the mask to breathe into a closed gas circuit. In that circuit, the oxygen is regenerated in the chamber 4. The flexible walled chamber or breathing bag 6 accommodates the expansion and contraction of the patient's lung.

In accordance with the present invention, the breathing bag 6 is suspended inside a rigid walled chamber 7 of considerably larger dimensions than the breathing bag 6. The interior of the breathing bag is connected to the conduit 5 and the chamber 7 outside the breathing bag 6 is supplied with gas through a conduit 8 at a pressure which is cyclically varied by apparatus constructed in accordance with the present invention. That apparatus includes a source of compressed air schematically indicated at 9. Any other gas under pressure greater than atmospheric may be used. There is no limitation on the selection of this actuating gas. Air is selected because it is cheap and convenient. From the source 9 of compressed air, the air flows through a special regulator valve 10 whose outlet pressure is closely controlled at a preselected value, for example 35 pounds per square inch gauge. Commercial pressure regulators are available which will hold such a pressure within an error of one or two-tenths of a pound, which is the error tolerance desired for this apparatus. At the regulator outlet there is connected a capillary bleed schematically indicated at 11. The purpose of this bleed is to maintain a continuing flow through the regulator even when the main discharge conduit is closed, thereby assisting the regulator in maintaining its outlet pressure within a narrow range. From the regulator 10 the air flows through a conduit 12 controlled by a valve 13 operated by a solenoid 14. The valve 13 is biased to closed position and is opened against its bias when the solenoid 14 is energized. The solenoid 14 is hereafter referred to as the charging solenoid. From the valve 13 the air flows through a conduit 15 and thence through a branch conduit 16 into one end of a cylinder 17. A piston 18 is slidable within the cylinder 17 by means of a piston rod 19. On the end of the rod 19 there is provided a rack 20 cooperating with a pinion 21 which may be rotated manually by means of a crank 22. The crank 22 fixes the position of the piston 18 in the cylinder 17 and thereby establishes the volume of the cylinder connected to conduit 16 at any desired value. The opposite end of the cylinder is vented to the atmosphere as indicated at 17a. The conduit 15 leads to a valve 23 biased to closed position and operable to open position by energization of a solenoid 24, hereinafter referred to as the inhale solenoid. Air passing through the valve 23 moves through a conduit 25, a throttle valve 26, a conduit 27, a flow regulating valve generally indicated at 28, a conduit 29, a flow meter and transducer 30 and a conduit 8 to the chamber 7. A conduit 31 branches from the conduit 29 and discharges through two parallel connected valves 32 and 33 to the atmosphere. The valves 32 and 33 are biased to closed position and are operable to open position by means of solenoids 34 and 35 respectively. The functions of the two valves 32 and 33 and their respective solenoids are the same. The two valves are provided rather than a single one for reasons of safety, to ensure that the system is vented to the atmosphere at the proper times. In

the further description of the apparatus of this invention, only one of these valves will be referred to, particularly the valve 32 and its solenoid 34 and it will be hereinafter termed the atmosphere solenoid.

Another branch conduit 36 leads from the conduit 29 through a valve 37 biased to closed position and operable to open position by a solenoid 38 hereinafter termed the forced exhalation solenoid. The valve 37 discharges through a vacuum pump 39 to the atmosphere.

Another branch conduit 40 leads from the conduit 29 through a pressure relief valve schematically indicated at 41, which discharges through a conduit 42 leading to an indicator housing 43. The housing 43 contains a light plastic ball, for example a ping pong ball, caged over the outlet of the conduit 42. When air flows through the conduit 42, the ball 44 is lifted in its cage so that the operator is advised that air is flowing through the relief valve 41.

The flow regulating valve 28 comprises a housing which is separated into three chambers 45, 46 and 47 by means of two diaphragms 48 and 49. The chamber 45 is connected to the conduit 15. The chamber 46 is connected through a throttle valve 50 which is manually adjustable and may be of the needle valve type, to the conduit 25. The chamber 47 is open to the conduit 29. The diaphragm 48 and smaller diaphragm 49 are attached to a central valve stem 51 which carries a valve 52 adapted to regulate the flow between the conduit 27 and the chamber 47. The valve 52 is biased toward closed position by a spring 53 whose tension may be regulated by means of a thumb screw 54. One or more additional springs 55 act on the valve 52 in the opposite direction. The balance between the springs 53 and 55 and the forces on diaphragms 48 and 49 determine the control of flow by the valve mechanism 28, dependent on the settings of the throttling valves 26 and 50.

The principal operating controls for determining the pressure cycle in the chamber 7 and hence the respiratory cycle of the patient on whom the mask 1 is mounted, are the four solenoids above described, namely the charging solenoid 14, the inhalation solenoid 24, the atmosphere solenoid 34 and the forced exhalation solenoid 38. The cycles of operation of these solenoids is determined by a group of relay circuits to be described below in connection with FIG. 5. The operation of the relay circuits is determined by certain manual controls to be described and also by certain controls responsive to conditions of flow or pressure related to the patient. The latter controls, hereinafter collectively termed the patient responsive controls, include a pressure switch generally indicated at 60 and comprising a flexible bellows 61 whose interior is connected to the conduit 15 and whose exterior is subject to atmospheric pressure. The bellows 61 operates a single contact and closes that contact as a predetermined low pressure slightly above atmospheric, for example, one pound per square inch.

Another of the patient responsive controls is hereinafter termed the assistor pressure switch and is shown at 62 in FIG. 1. The switch 62 comprises a bellows 63 whose interior is subject to the pressure in the chamber 7. It operates a single movable contact 64 between an upper stationary contact 64a and a lower stationary contact 64b. The assistor pressure switch is adjusted so that its contact finger 64 is between the stationary contacts when its internal and external pressures are balanced, i.e., both atmospheric. At a small value of positive pressure, for example two millimeters of mercury, the switch finger 64 engages contact 64a. At another small value of negative pressure, which may also be two millimeters of mercury, the contact finger 64 engages the stationary contact 64b. The positive and negative pressures referred to represent pressures within the chamber 7.

Another of the patient responsive controls is termed herein the integrator. This integrator is shown only dia-

5

grammatically in FIG. 1, being designated by the reference numeral 65. It is shown in greater detail in FIG. 2. Another patient responsive control is termed herein the exhalation sensor and is shown in FIG. 1 diagrammatically at 66, being illustrated more completely below in connection with FIG. 3.

Before starting a typical breathing cycle with the apparatus of the present invention, the cylinder 17 will be charged with air at the pressure determined by the regulator 10. This charging is accomplished by opening the valve 13 while the valve 23 remains closed.

After the charging is complete, the valve 13 is closed, trapping a predetermined volume of air in the cylinder 17 at a predetermined pressure.

It may therefore be seen that a fixed predetermined mass of air is trapped in the cylinder 17 and the conduits connected to it between the valves 13 and 23.

In carrying out a typical breathing cycle with the apparatus of the present invention, the valve 23 is opened by energization of the inhalation solenoid 24, allowing the air trapped in the cylinder 17 to flow through the valve 23 and through the throttle valve 26 and the flow regulating valve 28 into the conduit 29 and thence to the chamber 7. During this time the valves 37 and 32 are closed. This is termed the inhalation phase, since during this time the pressure is increasing in the chamber 7 and the breathing bag 6 is being contracted to force gas through the mask 1 into the patient's respiratory system. This inhalation phase continues until the pressure in the cylinder 17 drops to a predetermined low value established by the setting of the pressure switch 60. When that switch closes its contacts, the inhalation phase is terminated by relay circuits described below in connection with FIG. 5. The termination of the inhalation phase is characterized by the closing of the inhalation valve 23 and the opening of one or the other of the exhalation valves 32 and 37.

There follows an exhalation phase, in which one or the other of the valves 32 or 37 remain open. As described more completely below, one of these valves may be opened for part of the total exhalation phase and the other for the remainder of that phase. The exhalation phase is terminated, as described below, by the operation of the integrator 65. The integrator 65 receives a signal which is a measure of the rate of flow (or the pressure) of the air being received by the patient and integrates and stores a charge which measures the total quantity of air received by the patient during the inhalation phase. When the exhalation phase starts, the flow meter reverses its signal and the charge stored in the integrator is gradually reduced until it appears that the same volume of air (or a predetermined fraction thereof) has been breathed out by the patient, whereupon the integrator terminates the exhalation phase. Termination of the exhalation phase calls for a closing of the forced exhalation valve 37, if it is open.

The forced exhalation phase may be followed by a rest phase in which the atmosphere valve 32 is opened, but no effort is made by the apparatus to force the patient to inhale or exhale. In some cases, it may be desirable to use a phase known as a sealed rest phase, in which the atmosphere valve 32 is closed. The sealed rest phase is a potentially dangerous condition, since the patient cannot even breathe, and should only be used after taking careful precautions.

The flow meter and transducer 30 may be of any conventional sensitive type which will produce a direct current signal proportional to the rate of flow of gas to the meter. For example, it may be of the type utilizing a diaphragm subject to the difference in pressures at the entrance and throat of a Venturi, and carrying a sensitive electrical element which is stressed in accordance with variations in the diaphragm position produced by variations in the rate of flow. This resistance element may be connected in a direct current Wheatstone bridge

6

to produce the necessary electrical signal, which is then amplified by an amplifier diagrammatically indicated at 67. A pressure transducer 68 is connected in fluid communication with the chamber 7. Any conventional sensitive pressure transducer may be employed which will provide an electrical signal proportional to the pressure in the chamber 7. It may be of the same general nature as the flow meter and transducer 30, discussed above. The signal produced by the transducer 68 is amplified by an amplifier 69. A selector switch generally indicated at 70 enables the connection of the output of either of the amplifiers 67 and 69 to an integrator preamplifier 71 whose output is connected through wires 72 and 73 to a fractionating control shown more completely in FIG. 2, and whose output is in turn connected to the input of the integrator 65. The integrator 65 drives an output relay 75 which controls circuits to be described in detail in connection with FIG. 5.

The output of the amplifier 67 is also connected to the input of an exhalation sensor 66 described more completely below in FIG. 3.

INTEGRATOR AND FRACTIONATING CONTROL—FIG. 2

The output wires 72 and 73 of the integrator preamplifier are connected, as shown in FIG. 2, in parallel to a sign sensing circuit generally indicated at 76 and to an impedance matching network generally indicated at 77. The sign sensing circuit 76 and the impedance matching network 77 are part of the fractionating control 74 which also includes an inverting switch 78 and a fractionating relay 79. The output of the fractionating control 74 is fed through wires 80 and 81 to the integrator itself which is generally indicated at 85.

The fractionating control 76 comprises a twin triode 82. The grids of the two triodes are connected respectively through resistors 83 and 84 to the signal input wires 72 and 73, respectively. The anode of the one triode is connected through a load resistor 85 and a common load resistor 86 to the positive terminal of a source of unidirectional electrical energy shown as a battery 87 having a grounded center tap 88. The opposite terminal of battery 87 is connected through a balancing resistance 88 having an adjustable contact to the cathodes of the two triodes. The second triode has its anode connected through the winding of a relay 89 and a parallel capacitor 90 to the common load resistor 86 and thence to the positive terminal of battery 87.

If the incoming signal between the wires 72 and 73 has one polarity, for example, if the wire 72 is positive, then the upper triode 82 will be conductive and the load current will flow through resistor 85. On the other hand, the triode 82b will be substantially non-conductive and the relay winding 89 will be de-energized. However, if the incoming signal has the opposite polarity and the wire 73 is positive, then the triode 82b will be conductive and the relay 89 will be energized. The relay 89 operates a single contact 89a. Contact 82a cooperates with both back and front stationary contacts.

The reason for using the twin triode and the resistor 85 is to maintain a substantially balanced load on the power supply 87 under all conditions of operation. The apparatus of the present invention is particularly intended for use in a portable unit which may be readily plugged into a commercial power supply. The various direct current circuits are then supplied through rectifiers which are of limited capacity. In order to provide stable and consistent operation of all the circuits, and to minimize the voltage regulation of the rectifier circuits, it is desirable to maintain the loads on those rectifiers as nearly constant as possible. Similar techniques will be used in many of the circuits described below. However, since such techniques are not essential part of the present invention, they will not be emphasized in the following description.

The relay 89 controls the repeater relay 79 through the inverter switch 78 which enables the operator to select which polarity of signal he wants to cause energization of relay 79. In the position of switch 78 shown in the drawing, the relay winding 79 is energized through a circuit which may be traced from the positive terminal of battery 87 through wire 91, relay winding 79, contact 78a of switch 78 and the back contact of relay contact 89a to ground. Consequently, with the switch 78 in the position shown, relay 79 is energized when the incoming signal has a polarity such that wire 72 is positive, and relay 89 is de-energized. If the inverter switch 78 is thrown to its opposite position, then relay winding 79 is energized when relay 89 is also energized, i.e., when the incoming signal has a polarity such that wire 73 is positive. The energizing circuit for relay 79 may then be traced from the positive terminal of battery 87 through wire 91 to relay winding 79, contact 78a and its stationary front contact, contact 89a and its stationary front contact to ground. In either position of switch 78, there is provided a load balancing circuit through a resistor 92 and contact 78b of switch 78. This load balancing circuit is provided to maintain the load on power supply 87 constant, regardless of whether relay winding 79 is energized.

The fractionating network 77 is provided to change the proportion of the total signal appearing between wires 72 and 73 which is transferred through the fractionating control 74 to the integrator 65. The contacts a, b, c and d of relay 79 are arranged so that when that relay is energized, the full signal from input wires 72 and 73 is transferred through wires 80 and 81. The circuit may be traced from wire 72 through wire 93, front contact b of relay 79 and a coupling resistor 94 to wire 80. The corresponding circuit may be traced from wire 73 through wire 95, front contact d of relay 79 and a coupling resistor 96 to wire 81.

The fractionating network 77 comprises six resistors 101, 102, 103, 104, 105 and 106. These six resistors are connected in two parallel groups, respectively, 101, 102, 103, and 104, 105, 106 between the wires 93 and 95. In each group, the three resistors thereof are connected in series. The resistors 102 and 105 are fixed resistors of equal value, for example 4 megohms. The resistors 103 and 104 are also of equal value, for example 3 megohms and the resistors 101 and 106 are of equal value, for example 10 megohms. The resistors 101, 103, 104, and 106 have sliding contacts which are ganged for concurrent operation by a manual control knob 107.

When relay 79 is de-energized, the signal from wire 72 passes through wire 93, a portion of resistor 103, the slider of that resistor and a wire 108 to the back contact b of relay 79 and thence through resistor 94 to wire 80. In a similar fashion, the signal from wire 73 passes through wire 95, a portion of resistor 104, the sliding contact on that resistor, a wire 109 and back contact d of relay 79 through resistor 96 to wire 81. At the same time, contacts a and c serve to connect a resistor 110 between the wires 93 and 95 through an obvious circuit. The net effect is to maintain the impedance load on the amplifier 71 across wires 72 and 73 substantially constant in either condition of the relay 79. Furthermore, the various resistors of the network 77 maintain that load constant in substantially all positions of the control knob 107. Likewise, the input impedance to the integrator is substantially the same in either condition of the relay 79 and in all positions of the knob 97. Other known impedance matching and balancing techniques may be used to similar effect.

The signal appearing between wires 80 and 81 is a balanced signal, like that between wires 72 and 73. In other words, it is balanced with respect to ground so that if one wire is positive with respect to ground by a certain potential, the other wire is negative with respect to ground by an equal and opposite potential.

The signal from wires 80 and 81 is fed into an ampli-

fier schematically indicated at 111, having output terminals 112 and 113. A pair of capacitors 114 and 115 are connected between the respective output terminals 112 and 113 and the corresponding input wires 80 and 81. A reset relay 116 is provided having contacts a and b, which when closed shunt capacitors 115 and 114, respectively, thereby discharging them, and setting the integrated value registered by the integrator at zero. The potentials across the capacitors 114 and 115 are applied to the grids of a twin triode 117, comprising triode 117a and triode 117b. The twin triode 117 is supplied with energy from a suitable source of direct current shown as a battery 118, having a grounded center tap. The triode 117a has connected in its anode circuit a load resistor 119. The triode 117b has connected in its anode circuit the winding of an integrator output relay 120 and a parallel capacitor 121. The cathodes of the two triodes are connected through a resistor 123 having a sliding contact to the negative terminal of battery 117. A selector switch 122 is provided by which either half of the resistor 123 may be shunted. The sliding contact on resistor 123 and the switch 122 are set so that when the potentials across the capacitors 114 and 115 are zero, the triode 117b is conductive and the relay winding 120 is energized.

The polarities of the various signals should be selected so that the potential of output terminal 112 of the amplifier 111 is negative, while that of output terminal 113 is positive. When a signal is applied to the input terminals of amplifier 111, it is amplified therein and the amplified signal is fed back to the input terminals through the capacitors 114 and 115, thereby storing on those capacitors a charge which represents an integration of the input signal. As that charge increases, the signal applied to the grid of triode 117b becomes more negative and eventually a charge which represents an integration of the input signal cuts it off at some predetermined value of the integrated signal on capacitor 114. Whether the triode 117b is conductive or not, the charge continues to be integrated on the capacitors 114 and 115. The signal polarity will be chosen so that the capacitors 114 and 115 become charged during the inhalation phase described above in connection with FIG. 1. During the exhalation phase the direction of flow and the change in pressure is reversed so that the polarity of the signal applied to the amplifier input is reversed, the amplifier 111 continues to integrate the applied signal, but now its effect is subtractive with respect to the capacitors 114 and 115. When the charge on capacitor 114 reaches a predetermined low value, it will no longer be sufficient to hold the triode 117b off, and it will become conductive, energizing the relay winding 120. As explained below in connection with FIG. 5, the energization of relay winding 120 establishes circuits effective to terminate the exhalation phase.

While reference has been made above to the triode 117b being conductive or non-conductive, and it has been spoken of as being cut off under certain circumstances, it will be recognized that the important points in the operating characteristics of the triode are those when its output current becomes sufficient to energize the relay 120 to pick up its contacts and the point, usually somewhat different, where the output current drops sufficiently so that the winding 120, though previously energized, will no longer hold up its contacts a and b.

The fractionating control has the effect of making the exhalation phase stronger or weaker than the inhalation phase, and it may have the effect of withdrawing from the patient's respiratory tract more or less gas than was introduced therein during the inhalation phase. There may be an accompanying tendency to inflate or deflate the patient. However, the exhalation phase is always followed by a rest phase, in which the patient's respiratory tract may readjust itself to external pressures. The fractionating control is supplied for the purpose of investigating the responses of the patient's respiratory tract to such inflationary and deflationary forces. Investigation of such responses is important both in experimentation to de-

termine the nature of the functions of the respiratory tract, and in clinical investigation to compare such responses with the responses of normal respiratory tracts for purpose of diagnosis and the like. Utilization of these responses may also be important in some type of anesthesia, and in treatment of respiratory disorders.

EXHALATION SENSOR—FIG. 3

The exhalation sensor as shown in this figure is intended to control the shift of the apparatus between a free exhalation phase and a forced exhalation phase. The inhalation phase of the apparatus is always forced, but the exhalation phase may be either free or forced. When the system is operating in a free exhalation phase, it is desired to have a control device which will sense the rate of exhalation flow and will transfer the system from the free exhalation phase to the forced exhalation phase, if the patient is not producing sufficient exhalation by himself. The exhalation sensor illustrated in this figure supplies this need.

As there shown, the output signal from the flow amplifier 67 is transferred through two coupling resistors 124 and 125 to the grids of a twin triode 126, comprising an upper triode 126a and a lower triode 126b. A load resistor 127 is connected in the anode circuit of triode 126a, the winding of a relay 128 is connected in the anode circuit of triode 126b, in parallel with a capacitor 129. The signals reaching the grids of the twin triode 126 are coupled through a further pair of coupling resistors 130 and 131 to the grids of another twin triode 132, comprising an upper triode 132a and a lower triode 132b. The anode of triode 132a is connected in series with the winding of a relay 133 connected in parallel with a capacitor 134. The anode of triode 132b is connected in series with a resistor 135. The twin triodes 126 and 132 are supplied with electrical energy from a battery 136. The triodes 126a and 132b and the resistors 127 and 135 are load balancing elements only. The functional active parts of the circuit are the triodes 126b and 132a and the relays 128 and 133.

The flow amplifier produces a signal whose polarity indicates the direction of flow. The polarity of that signal is chosen so that during inhalation, the triode 132a is conductive and the relay 133 is picked up. The capacitor 137 becomes charged during the inhalation phase and is effective to hold the signal on the grid of triode 132a after the inhalation phase terminates until the charge has a chance to leak off through resistors 130 and 131. The triode 126b and relay 128, on the other hand, are arranged so that triode 126b is conductive and relay 128 is energized during an exhalation flow.

The relay 128 operates a single contact 128a which engages a stationary back contact. In a similar fashion, the relay 133 operates a contact 133a which engages a stationary back contact. The contacts 128a and 133a are connected in series in a circuit which is completed in FIG. 5. During the inhalation phase, and for a short time thereafter, this circuit is open at the contact 133a by virtue of the energization of relay 133. During a continued exhalation flow, the circuit is also open at contact 128a. However, after the initial part of the exhalation phase, contact 133a closes, and when and if thereafter the patient does not exhale at the predetermined rate, relay 128 will be de-energized, thereby closing contact 128a and establishing a circuit to be described below in connection with FIG. 5, which will shift the system to the forced exhalation phase.

A manually operable switch 138 is connected in parallel with the series connected contacts 138a and 133a. The closing of switch 138 eliminates the free exhalation phase so that the system goes from inhalation directly into forced exhalation. For that reason, the switch 128 is referred to as the "free exhalation out" switch.

VACUUM SELECTOR CIRCUIT—FIG. 4

This figure illustrates diagrammatically a circuit for controlling the vacuum pump which supplies sub-atmospheric pressure during the forced exhalation phase. As there shown, the motor is supplied through alternating current supply lines 139 and 140 through an auto-transformer winding 141 having a fixed tap 142 and an adjustable tap 143. A motor 144 drives the vacuum pump 39. One terminal of the motor is directly connected to supply line 140. The opposite terminal of the motor is connected to contact 145a of a relay 145. Contact 145a cooperates with a stationary back contact connected to fixed tap 142 and alternatively with a stationary front contact connected to the adjustable tap 143. The winding of relay 145 is connected in a circuit extending from supply line 139 to supply line 140, whose details are shown and described below in connection with FIG. 5. An alternative circuit for energizing relay 145 extends through an immediate vacuum switch 146. Energization of relay 145 shifts contact 145a from the back contact to the front contact.

The relay 145 is normally de-energized except during the forced exhalation phase. When relay 145 is de-energized, the motor 144 is energized from the tap 142, whose voltage is selected to be just sufficient to keep the motor 144 and pump 139 turning over, so that their speed can be increased to an effective value rapidly. Their speed may be so increased by energizing the relay 145, which transfers the motor to the higher voltage tap 143, whereupon the pump 39 produces a substantial negative pressure or vacuum.

RELAY CIRCUITS—FIG. 5

This figure illustrates diagrammatically the various relay circuits controlling the several phases of the respiratory cycles which may be produced in accordance with the invention and the several phase sequences which may be produced.

There are four phase relays, termed the inhalation relay 151, the free exhalation relay 152, the forced exhalation relay 153 and the rest relay 154. Each of these four relays controls contacts which establish circuits for energizing the solenoids 14, 24, 34 and 38 in a manner to establish the respective respiratory phases.

The sequence of energization of phase control relays is determined principally by the phase sequence relay 155, hereinafter termed the ventilate sequence relay or simply the ventilate relay, and relay 156, hereinafter termed the assist sequence relay or simply the assist relay.

In the normal operation of the apparatus, the selection of the ventilate or assist sequence is determined by the response of the patient (although subject to certain overriding controls by the operator). At the beginning of operation of the apparatus, either the ventilate or assist sequence must be selected by the operator, but thereafter the sequence selection is determined by the patient's response.

In order to describe the relay circuits, it is considered best to describe the operation of the various sequences and phases, tracing the circuits as they are established in the course of the operation.

Operation is begun by manually operating the start switch, generally indicated at 157, and having four contacts a, b, c and d. The start switch 157 is center-biased to the open position shown in the drawing and is movable therefrom either upwardly to a ventilate position where it picks up the ventilate sequence relay 155, or downwardly to an assist sequence position, wherein it energizes the assist sequence relay 156. For the purposes of illustration, it is assumed that the system is started by actuating the start switch 157 to its upper position to initiate the ventilate sequence.

When the start switch is moved to its ventilate position, its a contact closes a circuit for energizing the ventilate sequence relay 155. This circuit may be traced

11

from a positive power supply line 158, appearing at the lower left of the drawing, through the *a* contact of a standby switch, generally indicated at 159, a wire 160 which serves as the normal positive power supply line throughout most of the operation of the system, the winding of relay 155, wire 161 and the *a* contact of start switch 157, through a resistor 162 to grounded supply line 163.

The *b* contact of the start switch 157 completes an energizing circuit for the inhalation relay 151. This circuit may be traced from wire 160 through a wire 164, the winding of relay 151, wire 165, contact *b* of start switch 157 and resistor 162 to grounded supply line 163. Contacts *c* and *d* of the start switch 157 are open when it is in its ventilate position.

When the ventilate sequence relay 155 is picked up, its *a* contact closes an energizing circuit for a signal lamp 166 which may be a neon glow lamp. This circuit may be traced from a positive power supply line 167 through wires 168 and 169, contact *a* of relay 155 and the signal lamp 166 to the grounded supply line 163.

Contact *b* of relay 155 completes a holding circuit for that relay which may be traced from positive supply line 160 through the winding of relay 155, wire 161, a wire 170, contact *b* of relay 156, contact *b* of relay 155, and a wire 171 to grounded supply line 163. Contacts *c* and *d* of relay 155 are open when that relay is energized, and are connected in a holding circuit for relay 156, described below.

Contact *e* of relay 155 and contact *e* of inhalation phase control relay 151 cooperate at this time to set up an energizing circuit for the free exhalation relay 152. This circuit is placed under control of the pressure switch 60 and will be closed by that switch to energize the free exhalation relay, thereby terminating the inhalation phase and starting the free exhalation phase. This circuit may be traced from positive power supply line 160, wire 164, the winding of free exhalation relay 152, wire 172, contact *c* of the forced exhalation relay 153, a wire 173, contact *e* of relay 155, a wire 174, the switch 60, contact *e* of inhalation relay 151, and wires 175, 176 and 177 to the grounded supply line 163.

Contact *f* of relay 155 prepares a circuit for charging a capacitor 183 in the input circuit of a thyatron 191 controlling the rest circuit timing relay 178. The circuit for charging capacitor 183 may be traced from positive power supply line 167 through wire 168, a wire 179, contact *g* of rest phase relay 154, a resistor 180, wire 181, contact *f* of relay 155, variable resistor 182, and capacitor 183 to grounded supply line 163.

Contact *g* of relay 155 is connected in a circuit to be described below and utilized under certain conditions for transfer from the ventilator sequence to the assist sequence operation.

Summarizing the operation of the ventilator relay 155, its important functions so far as the initial phases of the ventilator sequence are concerned are the setting up of its own holding circuit and the preparation of a circuit for the free exhalation relay 152.

As mentioned above, the inhalation phase relay 151 was energized when the start switch moved to the ventilator position by a circuit previously traced and extending through contact *b* of the start switch.

Contact *a* of the inhalation relay 151 completes an obvious circuit for energizing a neon lamp signal 184, for indicating that the relay 151 is energized.

Contact *b* of relay 151 completes a holding circuit for that relay which may be traced from power supply line 160 through the winding of relay 151, wires 165 and 185, contact *b* of relay 151, contact *b* of relay 152, contact *b* of relay 153, contact *b* of the relay 154 and wire 186 to ground.

Contact *c* of inhalation phase relay 151 closes an energizing circuit for a power relay 187 controlling the inhalation valve solenoid 24. The energizing circuit for

12

relay 187 may be traced from power supply line 160 through the winding of relay 187, back contact *g* of relay 152, front contact *c* of relay 151 and wires 175, 176 and 177 to grounded power supply line 163.

Contact *d* of relay 151 completes an energizing circuit for the integrating reset relay 116. Part of this circuit appears in FIG. 2 and part in FIG. 5. Beginning in FIG. 2, there is shown a battery 188 which continually charges a capacitor 189 through a circuit including a resistor 190.

The energizing circuit for relay 116 may be traced from the ungrounded terminal of capacitor 189 through the winding of relay 116 and thence through a wire 191 (see also FIG. 5) through contact *d* of relay 151 and wires 175, 176 and 177 to grounded power supply line 163. Re-

lay 116 is only momentarily energized by the charge on capacitor 189, so that relay 116 resets the integrator at the beginning of the inhalation phase. The current thereafter supplied from battery 188 through resistor 190 is insufficient to hold the relay 116 picked up. Relay 116 is not again picked up until the capacitor 189 has had a chance to charge while the contact *d* of relay 151 is open.

Contact *e* of relay 151 prepares the pick-up circuit for the free exhalation relay 152. This circuit was described above in connection with contact *e* of the ventilator sequence control relay 155.

Contact *f* of the inhalation phase relay 151 is connected in the holding circuit for the rest phase relay 154, and is open when the inhalation relay is energized.

Contact *g* of relay 151 is connected in the energizing circuit of the timer relay 178, and is also open when relay 151 is energized.

Contact *h* of relay 151 provides a load balancing circuit through a resistor 147, corresponding to the load consisting of relay winding 178 and thyatron 191, which is energized through contact *g* when relay 151 is de-energized. This circuit may be traced from power supply line 167 through wires 168 and 192, a resistor 193, contact *h* of relay 151, resistor 147, and wires 194 and 195 to ground.

As noted above, energizing of inhalation phase relay 151 causes energization of the inhalation power relay 187. Contact *a* of relay 187 closes a circuit which shunts the capacitor 183 through a resistor 196. This circuit may be traced from the left-hand (ungrounded) terminal of the capacitor 183 through wire 197, contact *a* of relay 187, resistor 196 and wires 194 and 195 to ground.

Contact *b* of relay 187 completes an obvious energizing circuit for the inhalation valve solenoid 24, simply connecting it between power supply lines 148 and 198. The inhalation phase is established since valve 23 is opened by solenoid 24 and the gas previously trapped in the cylinder 17 now flows through the throttle valve 26 and the flow regulator valve 28 to the chamber 7, thereby compressing the breathing bag 6 and forcing gas from the closed gas circuit through the mask 1 to the patient. As the gas flows out of cylinder 17, the pressure therein gradually drops. This pressure is continuously measured by the pressure switch 60, which is set to close at some predetermined low value, for example one pound per square inch gauge. When the pressure switch 60 closes, it completes the circuit for energizing the free exhalation relay 152 which was traced in connection with contact *e* of ventilator relay 155. Energization of the free exhalation relay 152 breaks the holding circuit of the inhalation phase relay 151 at contact *b* of relay 152. De-energization of relay 151 opens the energizing circuit for relay 187 and hence for the solenoid 24, thereby terminating the inhalation phase. The energization of the free exhalation relay 152 completes a circuit for a charging power relay 199 and also completes an energizing circuit for an atmosphere power relay 220. The inhalation valve 23 being closed and the atmosphere valve 32 being opened by energization of the solenoid 34, the free exhalation phase is established.

The various circuits controlled by the free exhalation phase relay 152 are traced in detail below:

Contact *a* closes an obvious energizing circuit for a neon lamp signal 221, which indicates that the relay 152 is energized.

Contact *b* is opened when relay 152 is energized and as noted above is connected in the holding circuit for relay 151.

Contact *c* is connected in a holding circuit for the free exhalation relay. This circuit may be traced from power supply line 160 through wire 164, the winding of relay 152, back contact *c* of relay 153, front contact *c* of relay 152, and wires 176 and 177 to ground.

Contact *d* of relay 152 is connected in an energizing circuit for the forced exhalation phase relay 153, and prepares that circuit for completion by the exhalation sensor of FIG. 3. This circuit may be traced from power supply line 160 through wire 164, wire 222, back contact *c* of rest relay 154, the winding of relay 153, contact *d* of relay 152 and thence through a wire 223 to the exhalation sensor of FIG. 3 and through contacts 128*a* and 133*a* thereof to ground. As pointed out above in the description of the exhalation sensor, this circuit will be completed only if the movement of gas in the exhalation direction stops before the integrator indicates that the air breathed in by the patient has substantially all been exhaled.

Contact *e* of relay 152 completes an energizing circuit for the charging power relay 199. This circuit may be traced from power supply line 158 through wire 223*a*, the winding of relay 199, wire 224 to contact *e* of relay 152 and wire 225 to ground.

Contact *f* completes a circuit for the atmosphere valve relay 220. This circuit may be traced from power supply line 158 through wire 223*a*, the winding of relay 220, wire 226, contact *f* of relay 152 and wire 195 to ground.

Contact *g* of the relay 152 is connected in the circuit of the inhalation valve relay 187, and is open when the relay 152 is energized. This contact is provided to de-energize the relay 187 immediately upon energization of relay 152, without waiting for the relay 151 to drop its contacts.

The free exhalation phase being established by energization of relay 152, that phase will normally continue until it is terminated by the exhalation sensor of FIG. 3, closing the energizing circuit traced above through contact *d* of relay 152 for the winding of relay 153.

When relay 153 picks up its contacts, it maintains the circuits for the charging relay 199 and completes a circuit for energizing the vacuum valve relay 227. It also prepares a circuit for the picking up of the rest relay 154, and completes a circuit for the immediate vacuum relay 145 illustrated in FIG. 4. The holding circuit for relay 152 is broken at contact *c* of relay 153, so that the atmosphere valve relay 220 is de-energized, with resulting de-energization of solenoid 34 and closure of valve 32. With valve 32 closed and valve 37 open, and valve 23 remaining closed, the forced exhalation phase is established.

The circuits controlled by relay 153 may be traced individually as follows:

Contact *a* closes an obvious circuit for a neon lamp signal 228 which indicates that relay 153 is energized.

Contact *b* is connected in the holding circuit for relay 151, previously traced.

Contact *c* is connected in the holding circuit for relay 152, previously traced.

Contact *d* is connected in a holding circuit for the relay 153. This circuit may be traced from power supply line 160 through wires 164 and 222, back contact *c* of relay 154, winding of relay 153, contact *d* of relay 153 and wire 186 to ground.

Contact *e* of relay 153 is connected in the energizing circuit for relay 199, being in parallel with contact *e* of relay 152 in the circuit traced in connection with the latter contact.

Contact *f* of relay 153 completes an energizing circuit for vacuum valve relay 227. This circuit may be traced from power supply line 160 through the winding of relay 227, contact *f* of relay 153 and wire 195 to ground.

Contact *g* is closed when relay 153 is energized, and prepares an energizing circuit for rest relay 154. This circuit may be traced from power supply line 160 through wire 164, the winding of rest relay 154, contact *a* of the integrator output relay 120, contact *g* of relay 153 and wire 195 to ground.

Contact *h* of relay 153 completes an energizing circuit for the immediate vacuum relay 145 described above in connection with FIG. 4.

The forced exhalation phase continues until the integrator output relay picks up, showing that all or a predetermined proportion of the gas which flowed into the patient during the inhalation phase, has been exhaled.

When the integrator output relay picks up, its contact *a* completes the circuit for energizing the rest relay 154. This circuit was traced in connection with contact *g* of relay 153. Rest relay 154 in picking up maintains the circuit for the charging valve relay 199 and completes a circuit for the atmosphere valve relay 220. It also completes its own holding circuit and prepares a circuit for the timing relay 178, which will be energized after a predetermined time to terminate the rest phase and start the inhalation phase again.

The several circuits controlled by the rest relay may be traced as follows:

Contact *a* controls an obvious energizing circuit for a neon lamp signal 229, which indicates that the rest relay is energized. Contact *b* of the rest relay is connected in the holding circuit for relay 151, previously described.

Contact *c* of the rest relay is connected in the pick-up and holding circuits of the forced exhalation relay 153. These circuits have been previously traced.

Contact *d* is connected in parallel with contact *e* of relay 152 in the energizing circuit for the charging valve relay 199.

Contact *e* completes an energizing circuit for the atmosphere valve relay 220. This circuit may be traced from power supply line 158 through wire 223*a*, the winding of relay 220, wire 226, contact *e* of relay 154, a manually operable "sealed rest" switch 230 in the normal position shown in the drawing, and thence to grounded supply line 163. When the sealed rest switch is in its other position, this latter circuit, instead of being completed directly to ground, is completed through a neon signal lamp 231 in parallel with a capacitor 232 and in series with a resistor 233. The current flow through the winding of relay 220 is not then sufficient to pick up its contacts and thereby to energize the atmosphere valve solenoid 34 so that while the rest phase is otherwise established, the pneumatic circuit connected to the chamber 7 is closed, and the patient cannot breathe. This sealed rest phase is utilized to measure the pressure in the patient's lungs. The switch 230 is shown as being spring biased to its normal position. This is a dangerous phase to use, since the patient cannot breathe. The signal 231 is therefore provided to give a special warning that the sealed rest phase is in effect. There are described below in connection with FIG. 6, certain modifications of the circuit of FIG. 5 which may be utilized to establish the sealed rest phase for a predetermined time.

Contact *f* of the rest relay completes a holding circuit for the rest relay which may be traced from positive power supply line 160 through wire 164, the winding of relay 154, wires 234 and 235, back contact *f* of relay 151, front contact *f* of relay 154 and thence to ground.

Contact *g* of relay 154 completes a charging circuit for capacitor 183 connected in the input circuit of thyatron 191, which in turn has its output circuit connected in series with the winding of timing relay 178. The charging circuit for capacitor 183 may be traced from power supply line 167 through wires 168 and 179, contact *g* of

15

rest relay 154, resistor 180, wire 181, contact *f* of ventilator sequence relay 155 and variable resistor 182 to the positive terminal of capacitor 183, whose negative terminal is connected directly to ground. The circuit of thyatron 191 and relay winding 178 may be traced from power supply line 167 through wires 168 and 192, resistor 193, contact *g* of relay 151, wire 185, relay winding 178, and the anode-cathode circuit of thyatron 191 and wire 236 to ground. After the circuit through contact *g* of relay 154 is established for a time determined by the setting of variable resistor 182, the thyatron 191 becomes conductive, thereby energizing the relay 178, which closes its contacts. Contact *b* of relay 178 completes an energizing circuit for the inhalation relay 151, which may be traced from power supply line 160 through wire 164, the winding of relay 151, wires 237 and 238, and contact *b* of relay 178 to ground. As soon as this latter circuit is complete, the inhalation phase is re-established and the system is ready to go through the cycle again.

Contact *h* of the rest relay 154 is connected in a circuit utilized during transfer of the system from the ventilating sequence just described to the assist sequence. This circuit is described more completely below. For present purposes, it may be pointed out that the *h* contact of relay 154 connects the movable contact of the pressure switch 62 to ground through an "assistor out" switch 238, which is normally closed, but which may be manually opened.

ASSIST SEQUENCE

The assist sequence differs from the ventilator sequence in that the free exhalation and forced exhalation phases are omitted. The assist sequence then consists only of the inhalation phase and the rest phase. The system may be started in the assist sequence by moving the start switch downward from the position shown in the drawing. Contacts *c* and *d* then complete an energizing circuit for the assist relay 156, which may be traced from power supply line 160 through wires 239 and 240, contact *d* of the start switch, wire 241, the winding of relay 156, a wire 244, contact *c* of the start switch, through resistor 162 to grounded supply line 163. Note that contact *d* shunts a resistor 242 which would otherwise be connected in series with relay winding 156.

Energization of the assistor sequence relay results in preparing the pick-up circuits for the inhalation phase relay 151 and the rest phase relay 154 and also prepare a circuit for charging the timing capacitor 183 which determines the duration of the rest phase.

The specific circuits controlled by the assistor relay 156 may be traced as follows:

Contact *a* closes an obvious circuit for energizing a neon lamp signal 243 to indicate that the relay 156 is picked up.

Back contact *b* is connected in the holding circuit for ventilator sequence relay 155, which was traced in the description of that relay above.

Contact *c* closes a holding circuit for relay 156 which may be traced from positive power supply line 160 through wire 239, back contact *d* of relay 155, wire 241, the winding of relay 156, wire 244, back contact *c* of relay 155, front contact *c* of relay 156 and thence to ground at 163.

Contact *d* of relay 156 prepares a pick-up circuit for the inhalation phase relay 151. This circuit may be traced from power supply line 160 through wire 164, winding 151, wire 237, contact *d* of relay 156, the back stationary contact of pressure switch 62, the movable contact of that switch, a wire 245, the manual switch 238, contact *h* of rest phase relay 154 to ground.

Contact *e* of relay 156 prepares a pick-up circuit for the rest relay 154 which may be traced from power supply line 160 through the wire 164, the winding of relay 154, wire 234, contact *e* of relay 156, wire 174, pressure switch 60, contact *e* of relay 151 and wires 175, 176 and 177 to ground.

Contact *f* of relay 156 establishes a charging circuit for capacitor 183. This circuit is similar to that previously

16

traced including contact *f* of relay 155 and variable resistor 182, except that this circuit substitutes contact *f* of relay 156 for the corresponding contact of relay 155 and also substitutes a different variable resistor 246 for the variable resistor 182 of the previous circuit. The reason for the use of the different variable resistor is that the rest phase during the assist sequence must generally be longer than the rest phase during the ventilator sequence, because of the fact that during the rest sequence, the rest phase must take care of the entire exhalation.

During the assist sequence, the inhalation phase, which was established by the start switch is terminated by the pressure switch 60 completing the circuit for the rest relay 154. The rest phase is normally terminated by an effort of the patient to inhale, resulting in the switch 62 closing its back contact, and thereby completing the circuit for the inhalation phase relay traced above in connection with contact *d* of relay 156.

TRANSFER FROM VENTILATION TO ASSIST SEQUENCE

Regardless of the particular sequence selected originally as determined by the direction of operation of the start switch, the system will, unless manually overridden, check during each rest phase to determine whether the patient is making an effort to breathe. If the patient is making such an effort, the system will be immediately transferred to the assist sequence. If the patient makes no effort to breathe within a predetermined time, the system will start the inhalation phase again and will start it on the ventilation sequence rather than the assist sequence. Either one of these shifts from the ventilator sequence to the assist sequence or vice versa is controlled by the assistor pressure switch 62 in cooperation with the timing relay 178.

First assume that the system is operating in the ventilator sequence and that the rest phase begins by energization of the rest phase relay 154. In the ventilator sequence, the presence of the forced exhalation phase may result in there being a negative pressure in the system at the beginning of the rest phase. This negative pressure may give to the assistor pressure switch 62 a false indication that the patient is trying to inhale. The system is prevented from responding to this false indication by the use of a timing capacitor 250. The timing capacitor is controlled by contact *a* of the atmosphere valve relay 220. When that relay is de-energized, a resistor 251 is connected across the capacitor 250, for the purpose of discharging that capacitor. When the relay 220 is energized, as it always is during the rest phase, the capacitor 250 is charged through a circuit which may be traced from power supply line 160 through resistor 242, wire 241, a wire 252 capacitor 250, contact *a* of relay 220, a wire 253, manual switch 238 and contact *h* of rest relay 154 to ground. The capacitor 250 is at this time connected in parallel with the winding of the assistor relay 156, or may be so connected if the assistor pressure switch 62 is making either of its contacts. If the assistor pressure switch is closing either of its contacts at this time, it is an indication that the patient is trying to breathe, so that the system should be transferred to the assistor sequence rather than the ventilator sequence. If the switch 62 is making its back contact, the energizing circuit for the assistor relay may be traced from the positive terminal of capacitor 250 through wires 252 and 241, the winding of relay 156, wire 244, contact *g* of ventilator sequence relay 155 and thence to the back contact of pressure switch 62 and through that switch, wire 245, manual switch 238, contact *h* of rest relay 154 to ground. If the assistor pressure switch is making its front or upper contact, the circuit last traced may be followed through the winding of relay 156 and thence directly to the front contact of pressure switch 62 and thence along the circuit last traced to ground. It may be seen that completion of either of these circuits will energize the assistor sequence relay 156. If it is the back contact of the pressure switch 62 that is made, indicating

that the patient is trying to inhale, then the inhalation relay 151 is also energized to start an inhalation cycle. This energizing circuit for the relay 151 may be traced from power supply line 160 through wire 164, the winding of relay 151, wire 237, contact *d* of relay 156 and thence through the back contact of pressure switch 62 to wire 245, switch 238 and contact *h* of rest relay 154 to ground.

If during the ventilator sequence, the pressure switch 62 makes neither of its contacts during the rest phase (after the initial period required to charge capacitor 250), then the system stays on the ventilator sequence. The rest phase is terminated by the timing relay 178 as described above.

Now assume that the system is working on the assist sequence, with the relay 156 energized and that the rest phase starts. If the pressure switch 62 engages its front contact, indicating that the patient is trying to exhale, an additional circuit is completed for the assist relay 156, but since that relay is already energized, no change takes place in the system. If the pressure switch 62 engages its back contact, indicating that the patient is trying to inhale, then the circuit last traced is completed, the system stays in the assist sequence, and the inhalation phase is started immediately.

At the start of the rest phase in the assist sequence, the capacitor 183 starts to charge through the contact *f* of assist relay 156, as described above. If the pressure switch 62 is not actuated by an inhalation effort by the patient within a predetermined time, then the relay 178 is picked up. Contact *a* of that relay energizes the ventilator sequence relay 155 to shift the system to the ventilating sequence and contact *b* of relay 178 closes a circuit previously traced for energizing the inhalation phase relay 151. This circuit for energizing the ventilating sequence relay 155 may be traced from power supply line 160 through relay 155, wire 161, contact *a* of relay 178 and thence to ground. This circuit for energizing the inhalation phase relay 151 may be traced from power supply line 160 through wire 164, the winding of relay 151, wires 237 and 238 and contact *b* of relay 178 to ground.

MANUAL CONTROLS FOR MODIFYING SEQUENCE

In addition to the manual control switches previously mentioned, certain additional manual controls are provided for varying the sequences previously described.

The switch 138 appearing both in FIG. 5 and FIG. 3 shunts the output of the exhalation sensor and makes it appear to the system as though the patient were not trying to exhale. It may be used to modify the ventilator sequence by cutting out the free exhalation phase, making the system go directly from the inhalation phase to the forced exhalation phase. A manual switch 260 is connected in parallel with the contact *a* of the integrator output relay. This switch is normally open and may be manually closed to cut out the integrator from the system. By closing that switch, the ventilator sequence may be modified to cut out the forced exhalation phase, so that the system goes from free exhalation directly into rest.

A complete exhalation switch 261 is connected in series with the *a* contact of the charging valve power relay 199. These two switches in series shunt the contact *g* of the rest relay 154. The closing of this switch makes it appear to the system that exhalation is complete (regardless of whether the free exhalation or forced exhalation phase is being used). It starts the timer running on relay 178 so as to restore the inhalation phase after a predetermined time.

The assistor out switch 238 is utilized to open all the energizing circuits for the assistor relay 156, so that the system must then operate permanently on the ventilator sequence.

A standby switch is generally indicated at 159 in FIG. 5 and is normally in the position shown. In any emer-

gency situation, the standby switch may be actuated to its standby position. In that position, its contact *b* closes an obvious energizing circuit for the relay 220, its contact *c* closes an obvious energizing circuit for the relay 199. The energization of relay 220 connects the chamber 7 to the atmosphere, so that the patient is free to breathe normally. The energization of relay 199 results in opening the charging valve 13, so that the circuit will be ready to start again after the standby switch is returned to its normal position. The standby switch may be used under emergency conditions to make sure that the patient is being relieved of any pressure stress placed on him by the system. It also has a fourth contact *d* whose only function is to connect a resistor 262 across the power supply. The purpose of this switch is to maintain a nominal load on the power supply, either for warm-up purposes or to prevent a change in the potential of the power supply when the system is thrown back into operation.

It is usually desirable to connect the outputs of the flow amplifier 67, the pressure amplifier 69, and the integrator amplifier 111 to suitable recording devices so that the variations in those conditions may be studied. The recording devices have been omitted from the drawings of the present application in order to simplify those drawings.

While the circuit has been described as used in connection with a power supply of limited capacity, i.e., a capacity not much greater than the load of the system itself, it is obvious that it may be used in any convenient type of power supply. The use of a limited capacity power supply involves certain modifications of the system, for example, the resistor 262 described above, which would not be necessary with power supplies of higher capacity.

SEALED REST PHASE—FIG. 6

This figure illustrates certain modifications of the circuits of FIGS. 5 and 3 which are necessary in order to maintain the sealed rest phase for a predetermined time. In FIG. 6, those elements which correspond fully to their counterparts in previous figures have been given the same reference numerals and will not be further described.

Generally speaking, the circuit of FIG. 6 introduces a time delay between the energization of the free exhalation relay 152 and the energization of the relay 220 which actually starts the free exhalation phase. The purpose of this time delay is to give time for the measurement or recording of the pressure in the patient's lungs. Two time delay relays 270 and 271 are provided. The relay 270 may provide a short time delay, for example one-tenth of a second, for the purpose of measuring the instantaneous value of the lung pressure. The relay 271 has a somewhat longer time delay and is useful for making records of the time versus pressure characteristics of the lung. It has been found that while a normal lung with air trapped in it does not produce any substantial variation of pressure with time, certain types of disease in a lung will produce a condition where the pressure in the closed lung will decrease gradually with time due to relaxation or expansion of the lung walls. The system with the relay 271 in it may be used for recording such conditions for the purpose of diagnosis.

In FIG. 6, the contact *f* of free exhalation relay 152 is connected in series with a manual selector switch 275, having three positions. In the position shown in the drawing, the selector switch 275 completes the circuit to the atmosphere valve relay 220, and the operation is the same as FIG. 5. In the center position, the switch 275 completes an energizing circuit for the winding of time delay relay 270 and another energizing circuit for the winding of a relay 272. Relay 270 picks up its contacts a predetermined time after the relay winding is energized. Its *a* contact completes the energizing circuit for relay winding 220 and the system thereafter proceeds as

before. In the third or upper position of switch 275, the winding of a different time relay 271 is completed along with the energizing circuit for relay 272. The operation is the same as in the case of relay 270 except that the time delay period is longer.

The relay 272 has two contacts. Contact *a* completes a circuit for energizing the sealed rest signal 231 (see also FIG. 5). Contact *b* completes a circuit for putting a false signal into the grids of the exhalation sensor twin triodes 126. This signal maintains the relay 128 energized and prevents the exhalation sensor from transferring the system to the forced exhalation phase prematurely. Note that this signal is introduced into the exhalation sensor triodes immediately upon energization of the free exhalation relay, and that there is no time delay in its application to those triodes.

The back *b* contacts of the time delay relays are connected in series with the winding of relay 272, and serve to de-energize that relay as soon as the time delay period is over.

FLOW METER AND TRANSDUCER—FIGS. 7 TO 10

The flow meter and transducer 30 of FIG. 1 above must produce an output signal which is a substantially linear function of flow, in order that it may be integrated to a value which truly represents the total quantity of gas delivered to the patient. If the flow transducer is one of the general type described above, then its output signal will be an exponential function of flow rather than a linear function. That exponential function may be compensated electrically so that the signal supplied to the integrator is linear. Alternatively, a flow meter of the type described below may be employed which produces a pressure differential that is a linear function of the gas flow.

Referring to FIGS. 7 to 10, the device is designated by the numeral 200 and will be seen to comprise a composite block housing having upper and lower plates 202 and 204 between which a series of relatively thin plates *P* are sandwiched by a series of peripherally disposed through-bolts 206. The gas to be measured is delivered to and from the housing through inlet and outlet conduits 208 and 210 received in the bottom plate 204. Pressure taps 212 and 214 provide means for deriving corresponding pressure signals at spaced intervals in the path of flow of the gas.

As best seen in FIGS. 8 and 9, the laminated plates *P* have alternating plates *P'* in each of which is formed an elongated slot 216. The plates *P''* inserted between each of the successive plates *P'* form therewith elongated slot-like passages 216 extending, in parallel, longitudinally of the housing. The plates *P''* are provided with openings 208' and 210' which are of substantially the same width as the slots 216 and register therewith to form vertically extending manifold passages 208'' and 210'', FIG. 10, which communicate with each of the slot passages 216 at the opposite ends thereof. The passages 208'' and 210'' connect with the inlet 208 and outlet 210, respectively, thus effecting the flow of gas in parallel fluid paths through the several slot passages 216. The plates *P'* and *P''* are such that the slots 216 form substantially equal flow restrictions of capillary dimensions whereby the pressure drop of the fluid passing therethrough bears a linear relation to the flow thereof. It is a significant feature of this device that a pressure differential signal is obtained from spaced points intermediate the inlet and outlet of the capillary slot passages, which it will be seen excludes the pressure drops occurring at the end portions of the passages. Thus, the non-linear pressure drops resulting from the entrance and exit effects of the gas flowing through the restriction are eliminated from the desired linear pressure signal. Such a signal is obtained by means of the lateral slot passages 216*a* and 216*b*, formed in each of the plates *P'*, which open into the longitudinal slot passage 216 and terminate at their outer ends in openings

212' and 214'. Preferably the entrance openings of passages 216*a* and 216*b* are spaced inwardly from the ends of the slots 216 a distance at least as great as the width of the slots. The openings 212' and 214' together with the identical openings in plates *P''* combine to form continuous vertical passages 212'' and 214'', FIG. 10, which connects with the corresponding pressure taps 212 and 214. The pressure taps may be connected to any suitable calibrated, pressure-differential responsive scale, or other measuring instrument.

It will be seen from the above that the pressures obtained at the respective pressure taps provide a differential pressure signal having a substantially true linear relation to the flow which, when suitably calibrated, or taken with the known, total effective cross-sectional area of the capillary passages 216 permits the gas flow to be directly determined accurately over a wide range of flow rates. No correction or calibration for non-linearity is required. In a specific embodiment of this device the plates *P* were arranged to provide 40 capillary passages about 0.87'' by 0.016'' in cross-section, and four inches in length, with the pressure taps centrally disposed about 2 inches apart. This arrangement affords a means of extremely accurate flow determination without adjustment for non-linearity for flow rates of from zero to 200 liters per minute, which is of particular value in measuring respiratory gas flows, and affords less than 2% departure from linearity at flow rates as high as 800 liters per minute. It will be seen that this device is of particular value in the present apparatus wherein an integration is made of the pressure differential as a means of determining the volume of inhalation and/or exhalation. Such measurement is accurate and reliable for use in the control and operation of the respirator only as long as the resultant pressure differential signal is a true linear function.

It will be understood that the respirator apparatus hereinabove described may be applied directly to the respiratory track of the patient rather than applied through the medium of a closed breathing bag and regeneration chamber in which the patient's respiration is confined to a conventional closed respiration circuit as shown in FIG. 1. Thus, for example, if anesthetic gases are not being employed, breathing bag 6 may be eliminated. The anesthetic machine 4 may also be eliminated. Connection to the patient may then, for example, be made directly through a suitable intratracheal catheter or face mask attached to conduit 8. In such arrangement the pressure switch 62 and the pressure transducer 68 may be connected directly to the face mask and chamber 7 will be eliminated.

While we have shown and described a preferred embodiment of our invention, other modifications thereof will readily occur to those skilled in the art, and we therefore intend our invention to be limited only by the appended claims.

We claim:

1. Respirator apparatus comprising inhalation means for cyclically supplying to a patient repeated charges of gas of substantially equal mass, said inhalation means comprising a chamber of predetermined volume, means for cyclically charging said chamber with gas at a first predetermined pressure, and thereafter releasing gas from the charged chamber until the pressure therein falls to a second lower pressure, and means measured by the released gas for controlling the volume and pressure of a charge of gas flowing to the patient.

2. Respirator apparatus as defined in claim 1, including means for varying the volume of said chamber.

3. Respirator apparatus, comprising a face mask, an inhaled gas circuit connected to said face mask, a breathing bag having flexible walls, means connecting the interior of the bag to said inhaled gas circuit, means defining a rigid-walled chamber, means supporting said breathing bag within said chamber with the exterior of the breathing bag exposed to the pressure in the chamber,

and means for controlling the pressure in said chamber to produce an inhalation contraction of the breathing bag, said last-named means comprising a chamber of predetermined volume, means for supplying gas at a first predetermined pressure, valve means operable to connect said gas supply means to said predetermined volume chamber, second valve means operable to connect said predetermined volume chamber to said rigid-walled chamber, means for operating said first and second valve means cyclically first to connect said gas supply means to said predetermined volume chamber and raise the pressure therein to said predetermined pressure, and thereafter to disconnect said gas supply means from said predetermined volume chamber and to connect the latter to said rigid-walled chamber and to maintain said last-mentioned connection until the pressure in said rigid-walled chamber drops to a second predetermined pressure lower than said first pressure.

4. Respirator apparatus as defined in claim 3, comprising conduit means extending between said predetermined volume chamber and said rigid-walled chamber and including said second valve means, and throttle valve means in said conduit in series with said second valve means.

5. Respirator apparatus as defined in claim 4, including flow control valve means in said conduit responsive to the pressure drop across said throttle valve means.

6. Respirator apparatus comprising gas supply means, means to establish an inhalation phase in which gas is supplied from said supply means to a patient, means to establish a forced exhalation phase in which gas is withdrawn from a patient, means to produce an electrical signal indicative of the rate of flow of gas to and from the patient, means to integrate said signal, and means responsive to the integrated signal for terminating the forced exhalation phase.

7. Respirator apparatus as defined in claim 6, including fractionating means responsive to the direction of gas flow with respect to the patient for varying the proportion of the signal to be integrated.

8. Respirator apparatus as defined in claim 6, including means for resetting the integrator at zero at the beginning of each inhalation phase.

9. Respirator apparatus as defined in claim 6, including manually operable means for resetting the integrator at zero.

10. Respirator apparatus, comprising a face mask adapted to engage a patient's face, gas supply and withdrawal means connected to the mask for the supply to the patient of gas for inhalation and the withdrawal from the patient of exhaled gas, gas flow control means operatively connected to the gas supply and withdrawal means, said gas flow control means being operable selectively to establish either an inhalation phase in which gas is supplied from the supply means to the mask, a rest phase in which the flow of gas to or from the mask is inhibited, a free exhalation phase in which gas may flow from the mask to the withdrawal means, or a forced exhalation phase in which gas is forcibly withdrawn from the mask by action of the withdrawal means; an inhalation phase relay, a rest phase relay, a free exhalation phase relay, and a forced exhalation phase relay; means actuated by each phase relay when energized to operate the gas flow control means to establish the corresponding respiratory phase; a ventilator sequence control relay, and means including the sequence control relay and effective when the latter is energized to establish a repeated ventilator cycle in which said relays are energized one at a time in the following sequence: inhalation phase relay, free exhalation phase relay, forced exhalation phase relay, rest phase relay.

11. Respirator apparatus as defined in claim 10, including means comprising an assist sequence control relay effective when the latter relay is energized to establish a repeated assist cycle in which said inhalation phase relay

and said rest phase relay are energized alternately, means responsive to a condition indicative of a flow of gas from the mask during said rest phase and effective to energize the assist sequence control relay if such a flow takes place, and to energize the ventilator sequence control relay if no such flow takes place within a predetermined time after initiation of the rest phase.

12. Respirator apparatus, comprising a breathing device adapted to be placed in fluid communication with a patient's respiratory system, gas supply and withdrawal means connected to the breathing device for the supply to the patient of gas for inhalation and the withdrawal from the patient of exhaled gas, gas flow control means operatively connected to the gas supply and withdrawal means, said gas flow control means being operable selectively to establish any of at least three respiratory phases, each characterized by a distinctive condition of gas flow between the gas supply and withdrawal means and the breathing device, a plurality of phase relays corresponding in number to said phases, means controlled by said phase relays for operating said gas flow control means, said last-named means being effective when each relay is energized to establish a particular respiratory phase, selectively operable means for energizing the relays one at a time in any of a plurality of different sequences, said selectively operable means comprising a plurality of selectively energizable sequence control relays, and means controlled by each sequence control relay when it is energized and by the phase relays to preselect, while each phase relay is energized, the phase relay to be energized next in the sequence.

13. Respirator apparatus as defined in claim 12, including manually operable means for preventing energization of one of the sequence control relays.

14. Respirator apparatus, comprising a breathing device adapted to be placed in fluid communication with a patient's respiratory system, gas supply and withdrawal means connected to the breathing device for the supply to the patient of gas for inhalation and the withdrawal from the patient of exhaled gas, gas flow control means operatively connected to the gas supply and withdrawal means, said gas flow control means being operable selectively to establish an inhalation phase, an exhalation phase, or a sealed rest phase in which substantially no gas can flow to or from the patient, means for measuring the pressure in the patient's respiratory tract during said sealed rest phase, means for terminating the sealed rest phase after a predetermined time, and signal means for indicating that the sealed rest phase is established.

15. Respirator apparatus, comprising a breathing device adapted to be placed in fluid communication with a patient's respiratory system, gas supply and withdrawal means connected to the breathing device for the supply to the patient of gas for inhalation and the withdrawal from the patient of exhaled gas, gas flow control means operatively connected to the gas supply and withdrawal means, said gas flow control means being operable selectively to establish an inhalation phase, a free exhalation phase in which a patient is free to exhale by his own efforts, or a forced exhalation phase in which gas is forcibly withdrawn from the patient's respiratory tract, means for sensing cessation of exhalation, and means including said sensing means for terminating said free exhalation phase and initiating said forced exhalation phase.

16. Respirator apparatus as defined in claim 15, in which said means for sensing cessation of exhalation includes flow responsive means for producing an electrical signal of one polarity during inhalation and of the opposite polarity during exhalation, a pair of relay means, respectively responsive to signals of opposite polarities and connected to said signal producing means, time-controlled means for maintaining energization of one relay means for a predetermined time after inhalation has terminated, a pair of contacts, respectively controlled by said pair of relay means, and an electrical circuit including

23

both contacts in series and completed only when both
said relay means are deenergized.

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UNITED STATES PATENT OFFICE
CERTIFICATE OF CORRECTION

Patent No. 3,033,195

May 8, 1962

John Gilroy et al.

It is hereby certified that error appears in the above numbered patent requiring correction and that the said Letters Patent should read as corrected below:

Column 3, line 48, for "hereafter" read -- hereinafter --; column 6, line 25, for "wirese" read -- wires --; line 34, for "85" read -- 65 --; line 58, for "82a" read -- 89a --; line 73, after "not" insert -- an --; column 8, line 32, beginning with "nal. As that charge" strike out all to and including "of the integrated" in line 36, same column, and insert instead -- a charge which represents an integration of the input signal. As that charge increases, the signal applied to the grid of triode 117b becomes more negative and eventually cuts it off at some predetermined value of the integrated --; column 20, line 63, for "th" read -- the --.

Signed and sealed this 28th day of August 1962.

(SEAL)
Attest:

ESTON G. JOHNSON
Attesting Officer

DAVID L. LADD
Commissioner of Patents