FLUIDICS AND PNEUMATICS
PRINCIPLES AND APPLICATIONS IN ANAESTHESIA

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FLUIDIC AND PNEUMATIC TECHNOLOGY has matured rapidly in the last ten years and as a result ventilators exploiting this advance are currently being introduced (Ohio Anesthesia Ventilator, Monaghan Volume Ventilator). It seems appropriate, therefore, to introduce the principles and devices used in this new technology in the hope of increasing the anaesthetist's understanding and thence the usefulness of the machines which employ it. This review offers explanations of the operation and construction of some of the principal fluidic and pneumatic components, and presents designs for an automatic blood pressure cuff inflator, a blood pump and a high performance ventilator.

The concept of fluidics arose in the late 1950's and was partly the result of the introduction of a device called a fluid amplifier, nomenclature adopted from electronics. The fluid amplifier, shown diagramatically in Figure 1, is based upon the Coanda effect, named for its discoverer, a Romanian engineer, which describes the tendency for a jet of fluid (the fluid can be a liquid or a gas) issuing from a nozzle to adhere to the surface of the wall adjacent to it. As the fluid stream leaves the nozzle, it causes entrainment of gas around it and consequently creates a negative pressure along its sides. This negative pressure is greatest on the side of the stream adjacent to the nearest wall and deflects the jet towards that wall. As the stream approaches the wall the negative pressure declines and the stream locks onto the wall surface.

The Coanda effect can be seen as a corollary of the venturi effect or the result of Bernoulli's principle. In the case of the fluid amplifier, the wall to which the fluid jet adheres can be controlled by a lateral jet. Figure 1 shows that the momentary energizing of a lateral jet switches the main jet from one outlet to the other. Since the lateral control jet need be far less powerful than the main jet, the term fluid amplifier is applied. Although the fluid amplifier may be directly applied as a primary output device, such as in a ventilator or blood pump[1], nevertheless it is wasteful, in that the main jet must always flow whether switched to the outlet which is used or to the outlet which is not used.

Having introduced one of the early fluidic devices, it is now perhaps easier to grasp some of the basic principles and terminology. The term *fluidic device*, for example, should rigorously be applied only to devices in which there are no moving parts except the fluid movement itself. All devices with moving parts are *pneumatic*. The advantage of a fluidic device over a pneumatic device is, of course, its lack of moving parts and hence lack of wear and fatigue. However, the dis-

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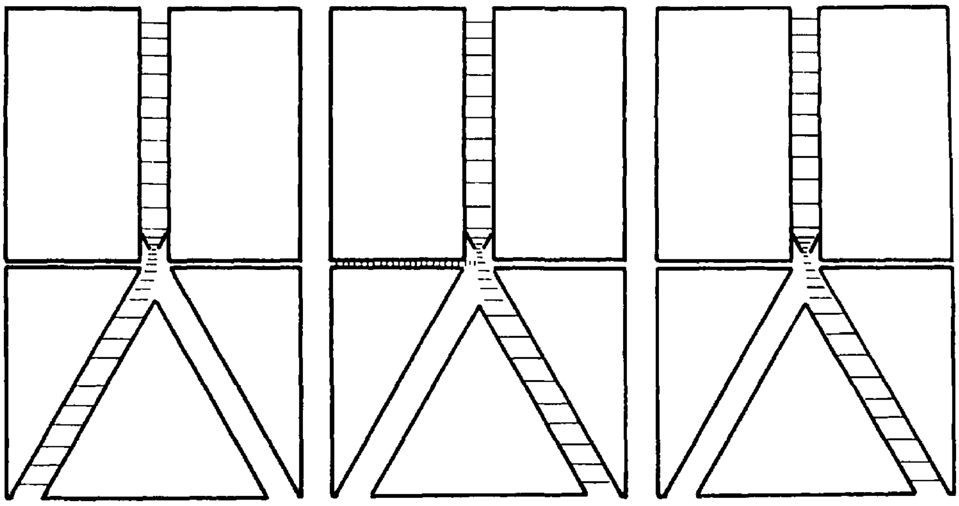


FIGURE 1. The fluid amplifier; showing the change in main flow direction induced by the momentary flow from a side port.

advantage is that the fluid stream must be continuous, so that flow is wasted and the energy loss is continuous. Pneumatic devices, on the other hand, can be made to turn on the flow only when necessary and so are less wasteful of energy. It is perhaps for these reasons that current technology tends to use fluidic devices for low pressure, low flow, control logic and then amplifies the output for the control of high pressures and high flows with pneumatic devices.

Fluidic Devices

About the simplest fluidic device is a flow resistor. There are two types, shown in cross-section in Figure 2 with their circuit symbols and pressure-flow characteristics. The top flow resistor is an orifice, with a nonlinear pressure loss versus flow curve, and the other is a capillary tube, with a linear pressure loss versus flow curve. A variable flow resistor is, of course, the familiar needle valve. This device is shown in cross-sectional view in Figure 3, with its circuit symbol and pressure-flow characteristics. It acts like a capillary tube when its flow resistance is high and like an orifice when its flow resistance is low.

Variants of flow resistance devices are also employed as limit switches or detectors for mechanical movement, with the proximity of an object changing the flow paths so as to change a flow resistance. Figure 4 shows two examples of limit switches. In the type at left, the insertion of an object increases the flow resistance of the device. In the type at right, the proximity of an object increases the flow resistance between the inflow and ambient enough to divert the flow to the output.

The more interesting fluidic devices, however, are the *logic devices*. Before explaining their operation it is necessary to explain some of the names applied to logic devices. Table I shows the names and functions of the more common logic devices and demonstrates their relation to the basic nor gate element.

There are several types of fluid logic devices commercially available. Those from

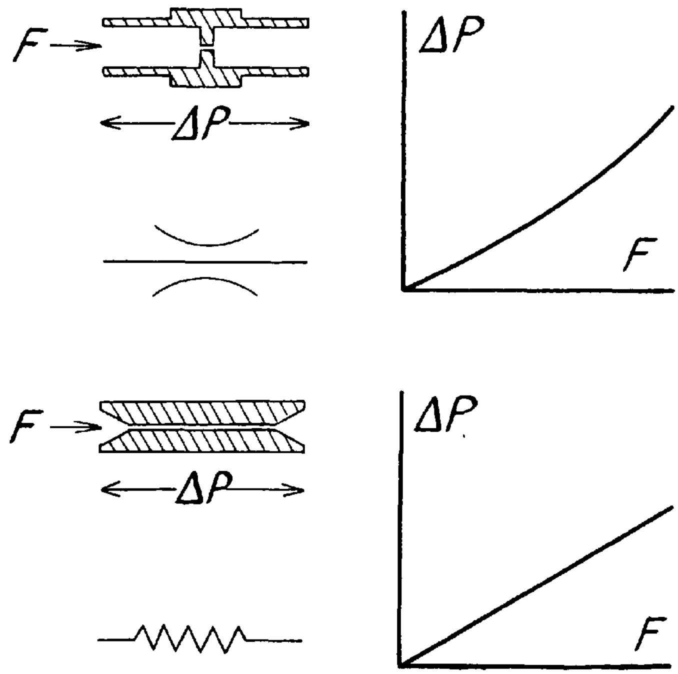


FIGURE 2. Flow resistances; top, the orifice, bottom, the capillary tube, with their circuit symbols and pressure-flow characteristics.

Corning use the principle of the fluid amplifier described earlier in Figure 1. Figure 5 shows symbols for the *or/nor, and/nand* and *flip-flop* logic functions.[2]

Johnson Controls fluid logic devices are based upon their summing impact modulator element outlined in Figure 6. Two opposing jets of air meet at an impact point and flow occurs radially from this point. Depending upon whichever jet is stronger the impact point is in either chamber A or B and therefore, either A or B output is energized. This device acts as a sensitive pressure comparator, and the same principle is used in the Johnson nor gate device.

The Maxam *nor gate*, called a turbulence amplifier, acts on the principle shown in Figure 7. A laminar flow jet impinges upon an outlet channel, energizing the output. If, however, the jet is made turbulent by the flow from one of four side jets then the turbulent flow exhausts to ambient and the output is turned off.

All of these devices operate using a continuous flow and are therefore designed as low pressure, low flow devices so as to minimize the constant energy loss entailed by their operation. In order to amplify the outputs to useful power levels without large energy losses a pneumatic interface valve is used.

PNEUMATIC DEVICES

Pneumatic devices used to be thought of only in terms of mechanical valves operated by a piston and cylinder arrangement such as the *spool valve* shown in Figure 8. Current technology, however, uses a diaphragm poppet design, shown in Figure 9, wherein the only moving parts are flexible rubber diaphragms (black

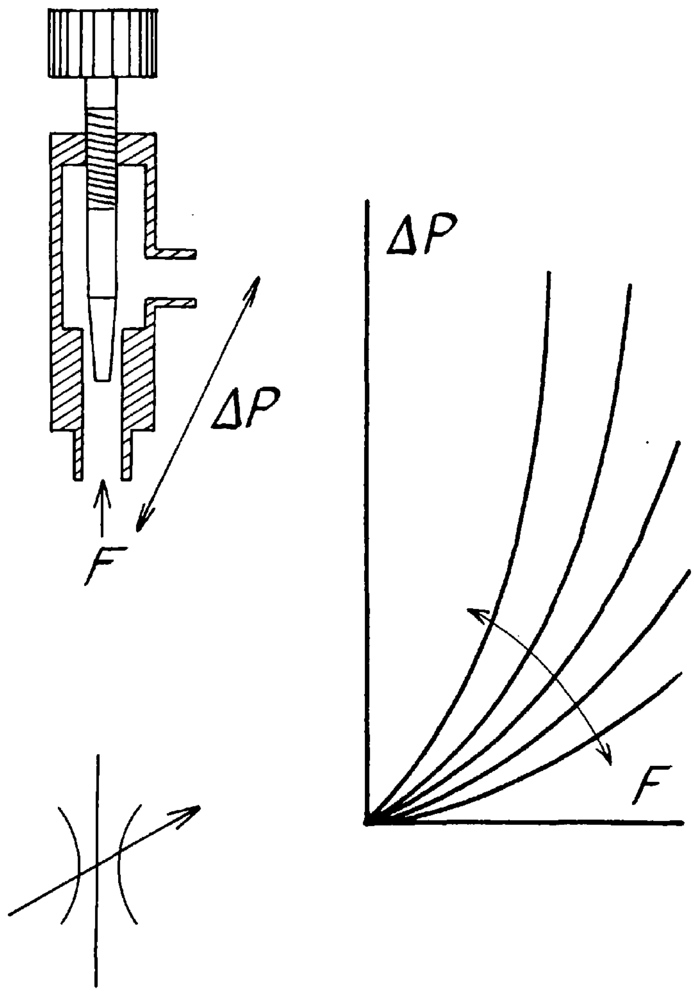


FIGURE 3. The variable flow resistance; with its circuit symbol and pressure-flow characteristic.

shading). Since these diaphragms are operated well below their elastic limit, they show almost no frictional wear or fatigue, and are rated for lives of many millions of operations. The use of a control diaphragm enables the interface valve to operate as a powerful amplifier; control pressures in the tenths of kPa range (cm of H_2O) can switch large flows in the range of hundreds of litres per minute at high pressures in the hundreds of kPa range (up to 100 psi). This control amplification entails a small continuous control flow, however, as for fluidic devices, but the high output flow only occurs when the valve is switched on.

The control diaphragm of the interface valve is also used as a separate device for low pressure applications. Figure 10 shows a diaphragm controlled valve in an amplifier circuit. In the valve-open position, as shown in Figure 10, flow from Ps, the supply, to ambient is resisted only by the orifice. The diaphragm experiences ambient pressure on one side and P_I on the other side. As P_I increases above ambient pressure, the diaphragm flexes and begins to restrict the flow from the

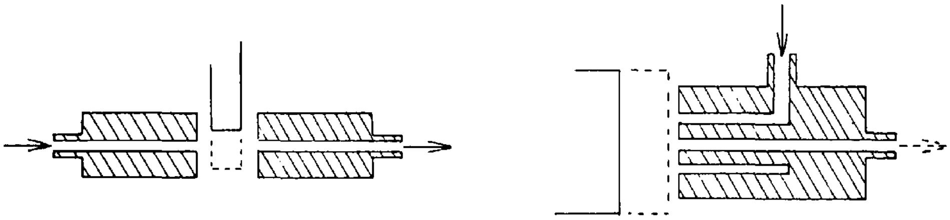
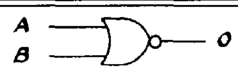
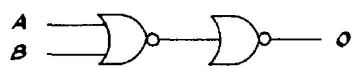
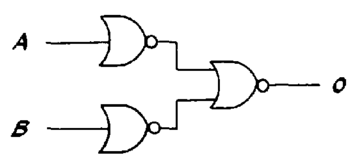
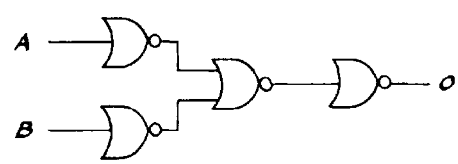
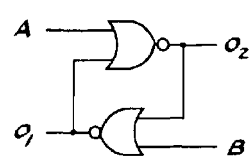


FIGURE 4. Limit or proximity switches; left, the interpositioning of a plate (dotted line) stops flow; right, the close approach of a surface (dotted line) starts the outflow.

TABLE I

LOGIC ELEMENTS: NAME, RELATION TO FUNDAMENTAL NOR GATE, AND LOGICAL FUNCTION

NOR GATE		A or B on → O off
OR GATE		A or B on → O on
AND GATE		A and B on → O on
NAND GATE		A and B on → O off
FLIP-FLOP or BISTABLE		O_1 on after A on O_2 on after B on

nozzle causing the output pressure, Po, to increase. Within a very short range of P_I the diaphragm occludes the nozzle completely and Po becomes equal to Ps. The graph in Figure 10 shows the input-output characteristic for the diaphragm controlled valve operated in an amplifier circuit. When a spring is used instead of a control pressure the diaphragm controlled flow device becomes the familiar regulator, as shown in Figure 11.

There are numerous other pneumatic devices of specialized functions including push buttons and on-off switches and at least one other should be noted, and that

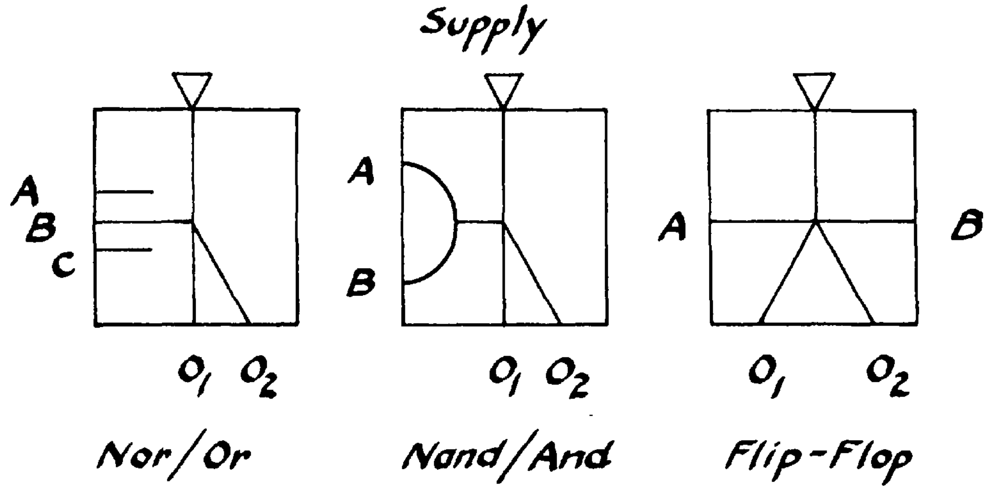


FIGURE 5. Symbols for the Corning logic devices; based upon the principle of the fluid amplifier. Left, nor/or, depending upon which output is chosen; centre, Nand/and, depending upon which output is chosen; and right, flip-flop or bistable.

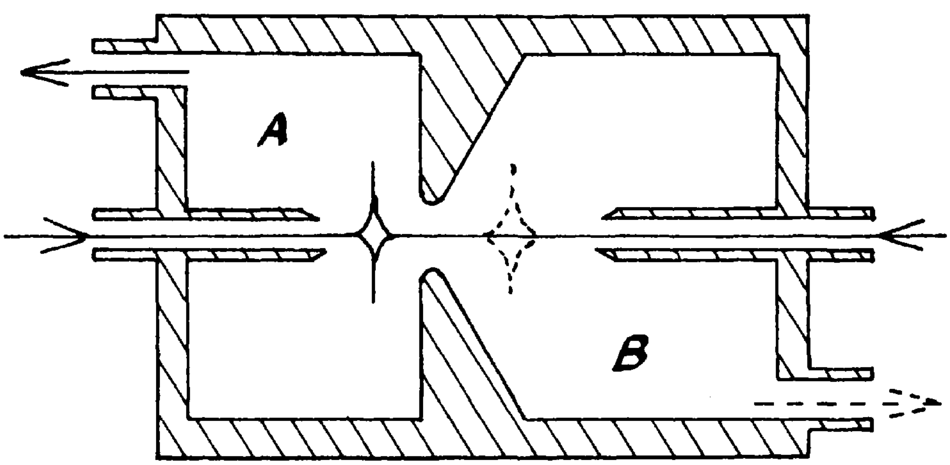


FIGURE 6. Johnson Controls summing impact modulator; when pressure at the left port becomes stronger than pressure at the right port, the flow impact point moves (dotted line) activating output B instead of A.

is the one-way valve. Anaesthetists are already familiar with the various ways of making a check valve or a one-way valve (the duck-bill type, the ball type, the spring-loaded disc type) so it is only necessary to record that one-way valves for both low pressure and high pressure applications are available for fluidic-pneumatic circuits.

Having described and explained some of the components used in fluidic and pneumatic circuits it remains to illustrate how they may be assembled into circuits which provide useful and complex functions. The next three sections describe, in order of increasing complexity, a simple automatic blood pressure cuff inflator, a blood pump, and a ventilator.

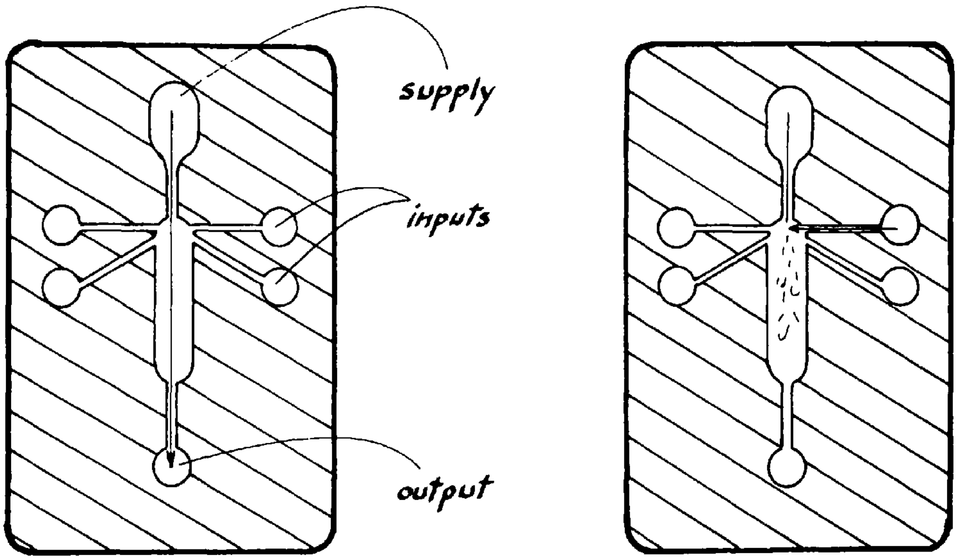


FIGURE 7. The Maxam nor gate; the output is present unless the laminar source jet is broken up by any of 4 input jets.

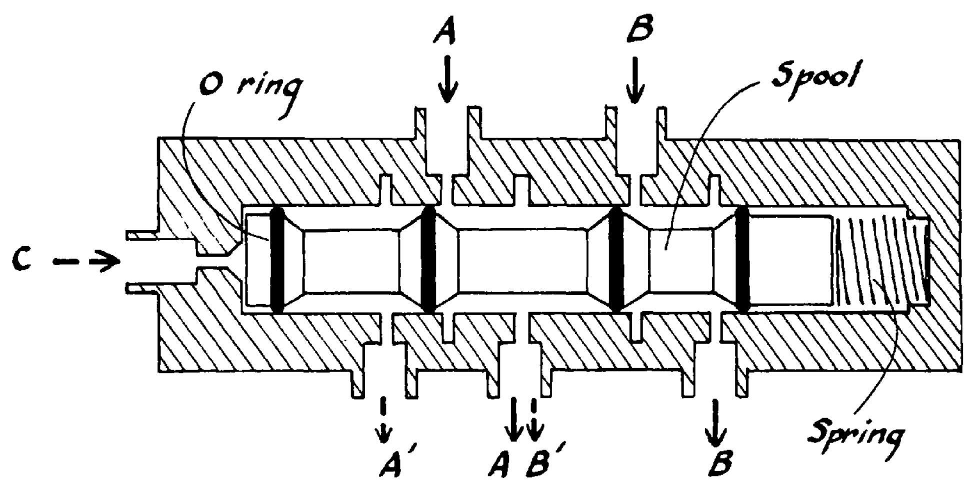


FIGURE 8. The spool valve; when pressure C acts upon the spool as a piston within the cylinder it moves the spool to the right changing the flow paths from A-A and B-B to A-A[1] and B-B[1]. The spring returns the spool to its original (shown) position when pressure at C stops.

AN AUTOMATIC BLOOD PRESSURE CUFF INFLATOR

Figure 12 shows a simple circuit that was used to inflate a blood pressure cuff cyclically. The cuff pressure and a superimposed Korotkoff sound transducer were continuously recorded, enabling blood pressure to be sampled continuously and automatically. The circuit operates as follows:

As the high pressure supply is turned on, a miniature needle valve (Johnson variable restrictor FCV-20) adjusts the flow through the nozzle of a diaphragm controlled valve (Johnson diaphragm amplifier, FPC-20) and through the orifice

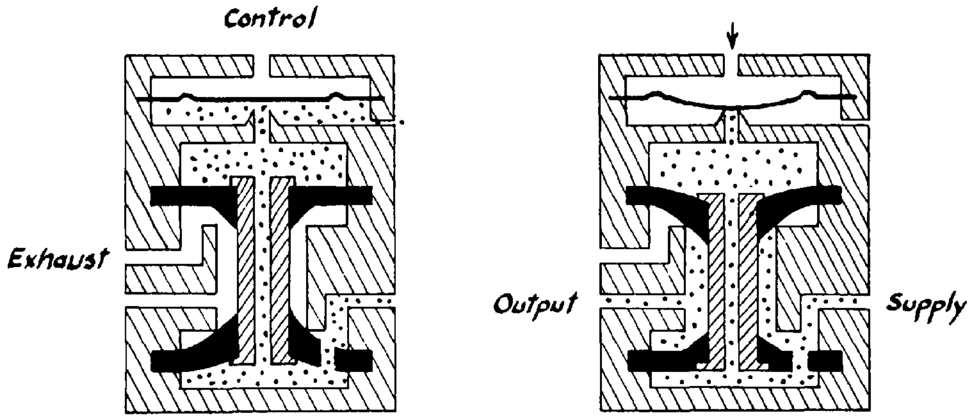


FIGURE 9. The interface valve; right, control diaphragm relaxed, output connected to exhaust; left, control diaphragm depressed, output disconnected from exhaust and connected to input.

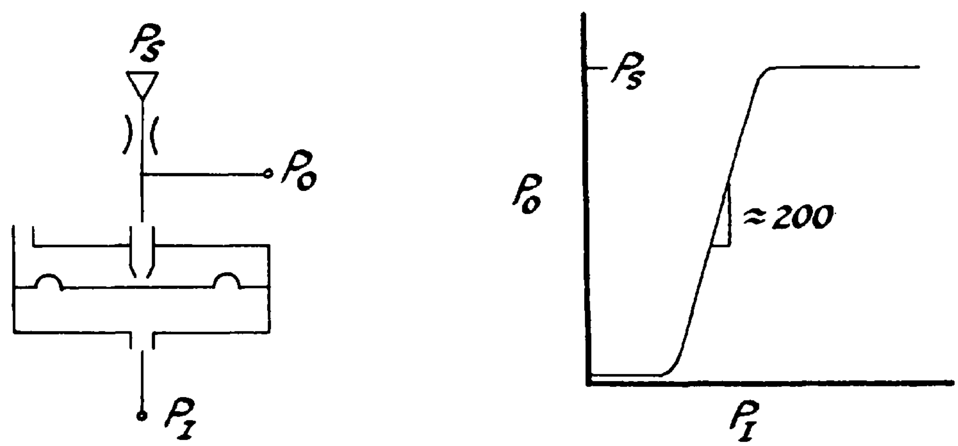


FIGURE 10. The diaphragm controlled valve in an amplifier circuit, with the input-output pressure characteristic.

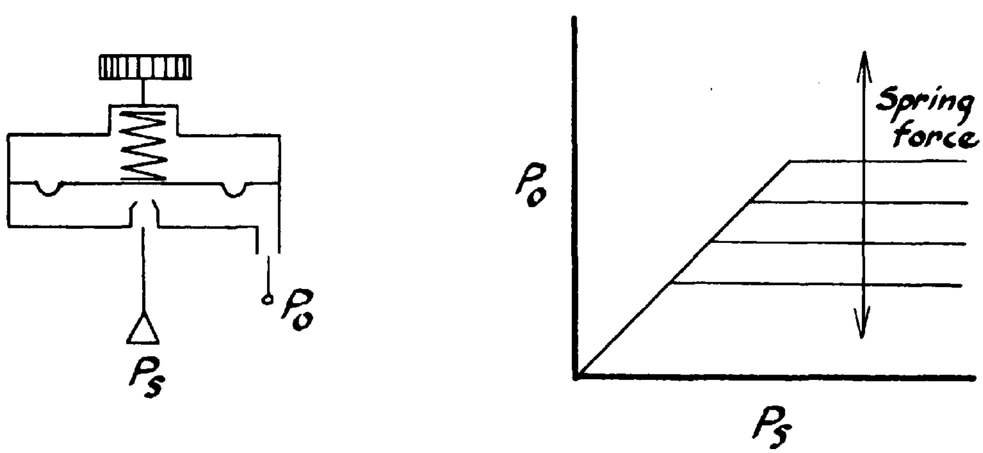


FIGURE 11. The diaphragm controlled valve in a regulator configuration, with the input-output pressure characteristic.

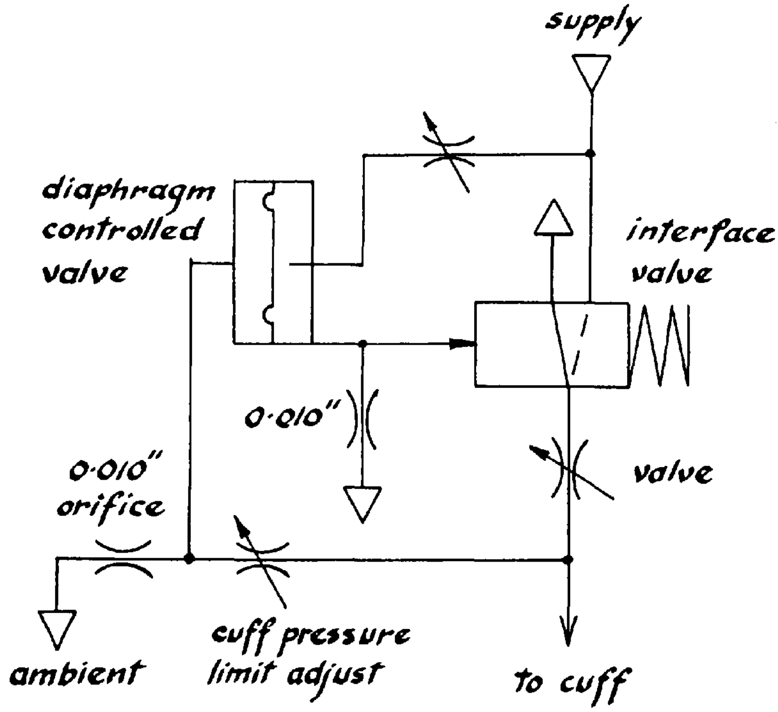


FIGURE 12. Circuit diagram for an automatic blood pressure cuff inflator; see text for details.

flow resistor (Johnson) to ambient air so that the pressure at the interface valve (Clippard 2010) control port is just sufficient to switch the high pressure supply through the interface valve. It should be mentioned that the symbol for this interface valve used here is not the conventional symbol but a modified version more illustrative of the valve's action. A conventional needle valve is used to adjust the rate of flow to the cuff. The pressure in the cuff causes a flow through another miniature needle valve and an orifice flow resistor to ambient air, thereby establishing a pressure at the control port of the diaphragm controlled valve which is less than cuff pressure. This pressure builds up as the cuff pressure increases until it is greater than the pressure on the outlet side of the diaphragm controlled valve, at which point the flow through the diaphragm controlled valve ceases and the interface valve shuts off. The cuff deflates through the needle valve and the interface valve exhaust port. As the cuff pressure declines so does the pressure at the control port of the diaphragm controlled valve. However, since the pressure on the outlet side of the diaphragm is now close to ambient, the diaphragm controlled valve remains shut off until cuff pressure has declined almost to zero. At this point the control port pressure is too low to prevent the flow through the diaphragm controlled valve, and pressure is again established at the interface valve control port and the cycle begins again.

A Blood Pump

The circuit for a blood pump for extra-corporeal circulation is shown in Figure 13.

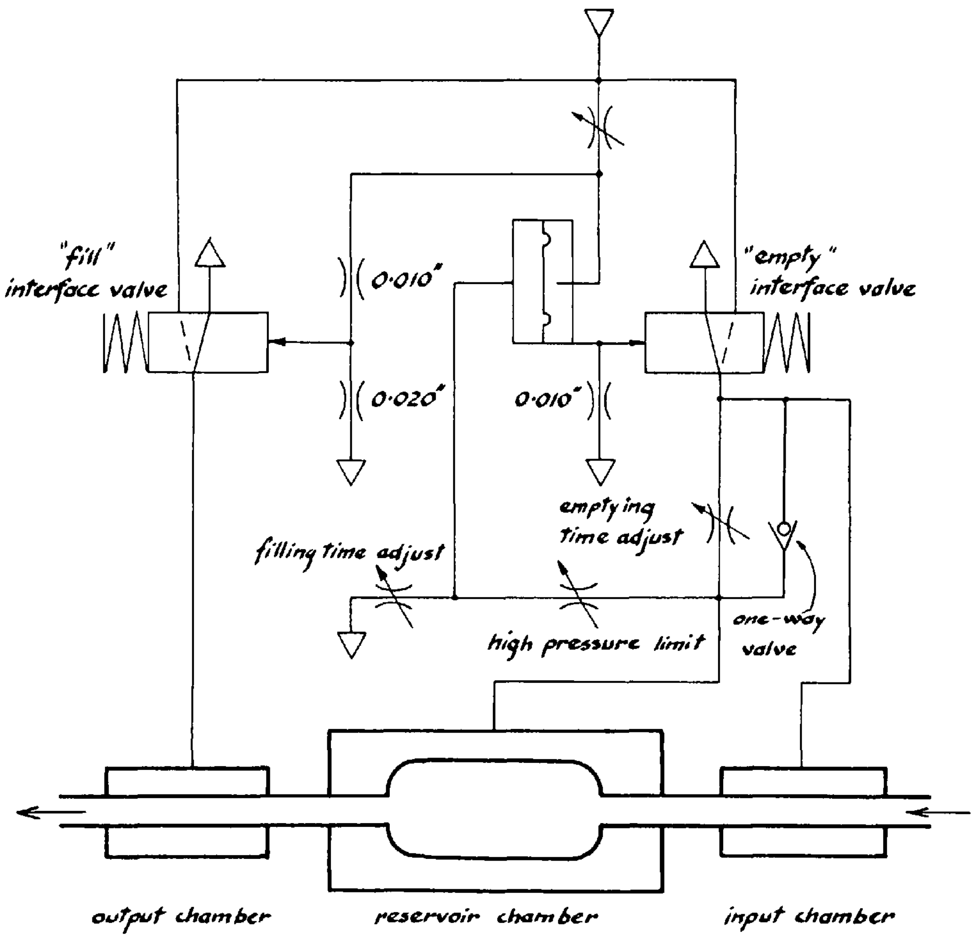


FIGURE 13. Circuit diagram for a blood pump; see text for details.

The basic pumping components are three enclosed chambers. The two smaller chambers act as one way valves and the larger chamber contains a reservoir bag of silicone rubber. During the filling phase the output chamber, through which thin-walled silicone rubber tubing passes, is pressurized from the high pressure source, effectively pinching off the tubing. Blood flows into the central reservoir bag through the open input chamber tubing. During the emptying phase, the pressure on the output chamber is released, opening the tubing at the same time as pressure is applied to the input chamber, pinching off the thin-walled input tubing. Pressure is slowly applied to the central chamber emptying the reservoir bag. The cycle then repeats. The pneumatic-fluidic circuit for applying the controlled pressures to the three chambers is an extension of the cuff inflator circuit, and operates as follows:

Pressure is applied from the source to the nozzle of a diaphragm controlled valve (Johnson FPC-20) via a miniature needle valve (Johnson FCV-20). This valve is adjusted so that when the diaphragm controlled valve is open, sufficient pressure is developed to turn on the "empty" interface valve (Clippard 2010) but not the "fill" interface valve; and when the diaphragm controlled valve is closed (and the "empty" interface valve is off) sufficient pressure is developed to turn on the "fill" interface valve.

The pump starts with the "empty" interface valve on, so that high pressure is applied directly to the inlet chamber and to the reservoir chamber via a needle valve (emptying time adjust). The flow to the reservoir chamber determines the rate of build up of pressure in the reservoir chamber and therefore the rate at which it empties. The pressure in the reservoir chamber is also applied to a miniature needle valve (high pressure limit) (Johnson FCV-20) and a small flow via this valve reaches ambient through another miniature needle valve (filling time adjust). The effect of this string of flow resistances is to apply a low pressure, proportional to the reservoir chamber pressure, to the control port of the diaphragm controlled valve. As the reservoir chamber pressure rises, so does the control port pressure and eventually it reaches a value sufficient to turn off the diaphragm controlled valve. This action turns off the "empty" interface valve and turns on the "fill" interface valve. Pressure in the reservoir and inlet chambers rapidly exhausts to ambient via the one-way valve (Clippard MCV-1) and the "empty" interface valve. The pressure at the control port of the diaphragm controlled valve exhausts more slowly to ambient via both the filling time miniature valve and the high pressure limit miniature valve. For this reason these controls interact. When the diaphragm controlled valve opens, the fill phase ends and the cycle begins again.

The pump has the advantages of a limited output pressure, of no inlet suction pressure, and of no complex one-way valves for blood, as well as being fully adjustable for emptying and filling times.

A Fluidic-Pneumatic Ventilator

Figure 14 shows the complete ventilator housed in an aluminium case. The ventilator is an all purpose instrument designed for the critical care area but also applicable to paediatric, anaesthetic and emergency use. It is operated entirely by gas pressure > 200 kPa (30 psi), and this driving gas is used to ventilate the lungs.

The ventilator is basically a time cycled constant flow generator thereby providing a constant volume. Inspiratory time (T_I) and expiratory time (T_E) may be independently varied, from 0.5 to 4.0 seconds for inspiratory time and from 1 to 7.5 seconds for expiratory time. Flow is continuously variable from 0 to 600 millilitres per second. Each of these controls have calibratable digital readouts. A high pressure alarm and cut off cycles the ventilator from inspiration to expiration whenever the output pressure exceeds a set pressure limit (variable between 1.5 to 6 kPa or from 15 to 60 centimetres of water). In addition, a red indicator flashes and a high pitched audible tone sounds.

A patient trigger control is also included. When output pressure decreases below a pre-set value (variable between −0.4 and +0.8 kPa or from −4 to +8 centimetres of water) during expiration the ventilator is cycled to inspiration and a green indicator flashes. The patient is protected against ventilator malfunction or tubing disconnection by a low pressure alarm (low pitched audible tone) which sounds if outlet pressure does not exceed 0.5 kPa or 5 centimetres of water during inspiration.

When used as a long term support ventilator the unit would require an oxygen

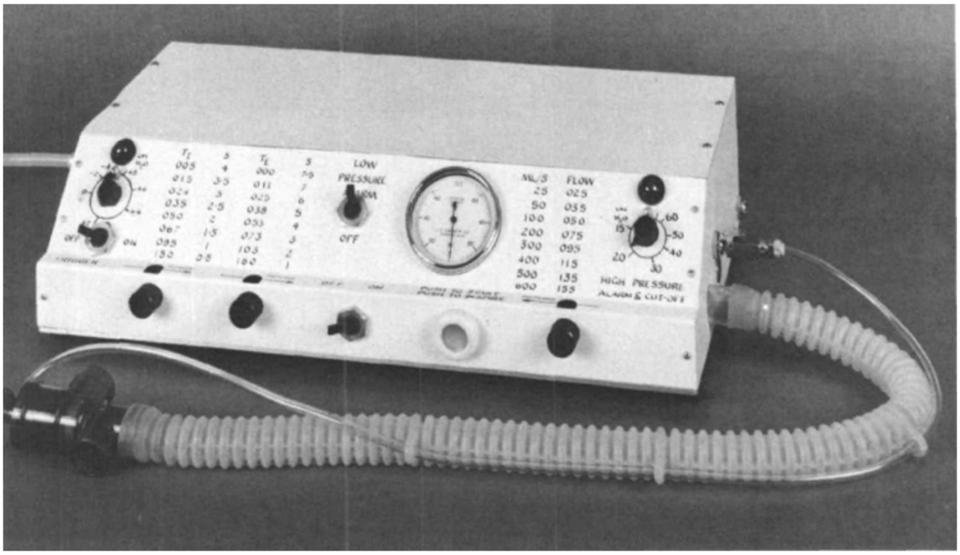


FIGURE 14. The ventilator, external appearance.

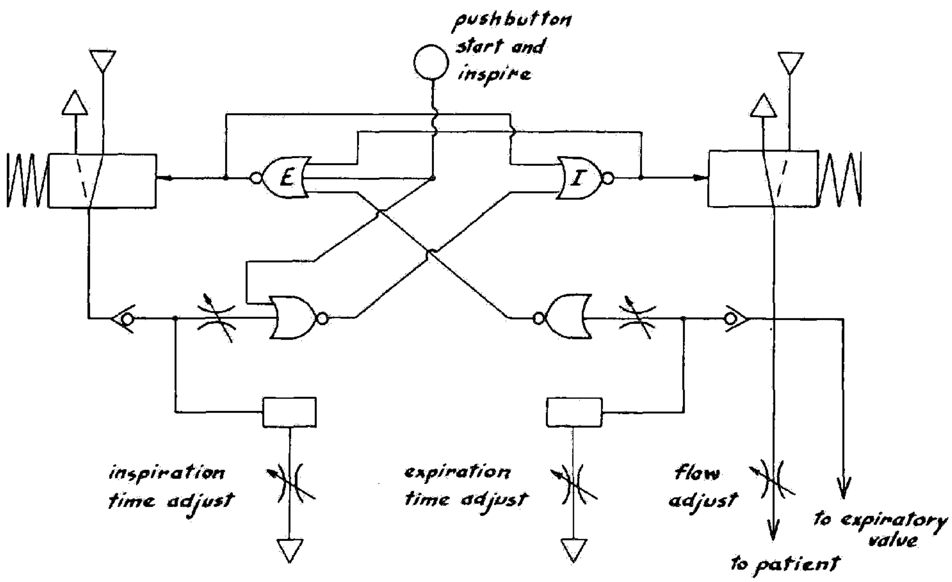


FIGURE 15. Circuit diagram for the basic oscillator of the ventilator; see text for details.

blender for the supply gas, a humidifier and a unit for positive end-expiratory pressure. The ventilator, excluding packaging, is made from standard industrial parts. Its performance equals or exceeds current ventilators of all types. It is small in size and light in weight.

Figure 15 shows the basic oscillator circuit of the ventilator. The I and E nor gates (Maxam) are connected to form a flip-flop circuit, each driving an interface valve. The timing circuits for inspiration time and expiration time are identical and operate similarly. For example, during the first few milliseconds of expiration,

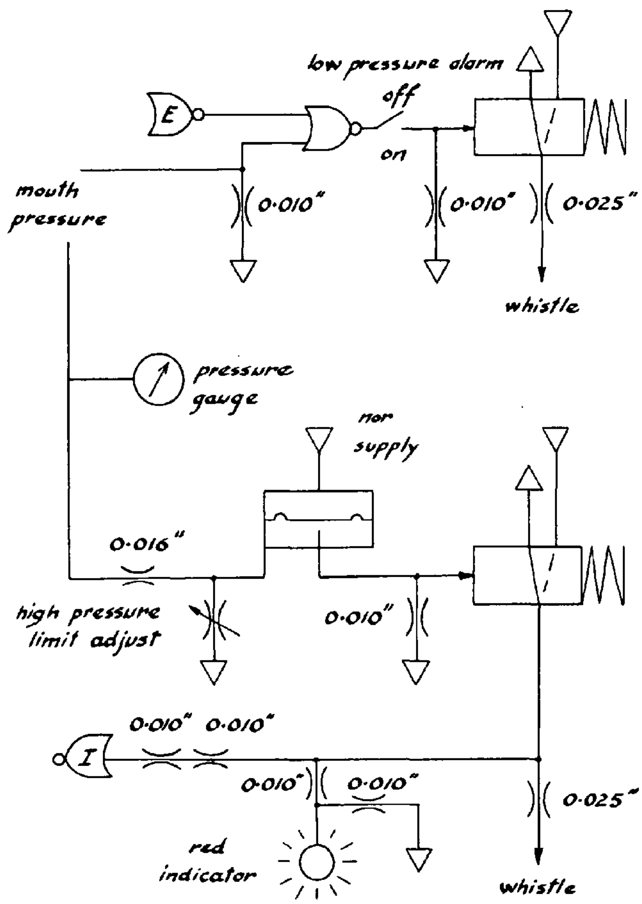


FIGURE 16. Circuit diagram for low and high pressure alarms for the ventilator; see text for details.

a high pressure (175 kPa or 25 psi) is applied via a one-way valve (Clippard MCV-2) to a fixed volume (shown as a rectangular box above the inspiration time adjust valve in figure 15) made from a length of tubing. Flow from this fixed volume to ambient is via both an inspiration time adjust valve (Matheson N.R.S. high accuracy valve with numerical counter handle) and via a miniature needle valve (Johnson FCV-20) where it holds the output of a nor gate off. The high pressure is maintained in the fixed volume until expiration ceases. At that time the pressure in the fixed volume begins to decline. The rate of decline is controlled entirely by the inspiration time adjust valve. When the pressure at the nor gate input has declined to near zero, the nor gate output turns on, switching the I nor gate off and terminating inspiration. Expiration time is controlled independently of inspiration time in the same manner.

During inspiration, the high pressure is supplied to the patient via an interface valve and a high flow resistance flow adjust valve (Matheson N.R.S. high accuracy valve with numerical counter handle), and also to the patient's expiratory valve. Because the pressure at the patient's mouth is insignificant when compared to 175 kPa or 25 psi, the pressure drop across the flow adjust valve is constant and, there-

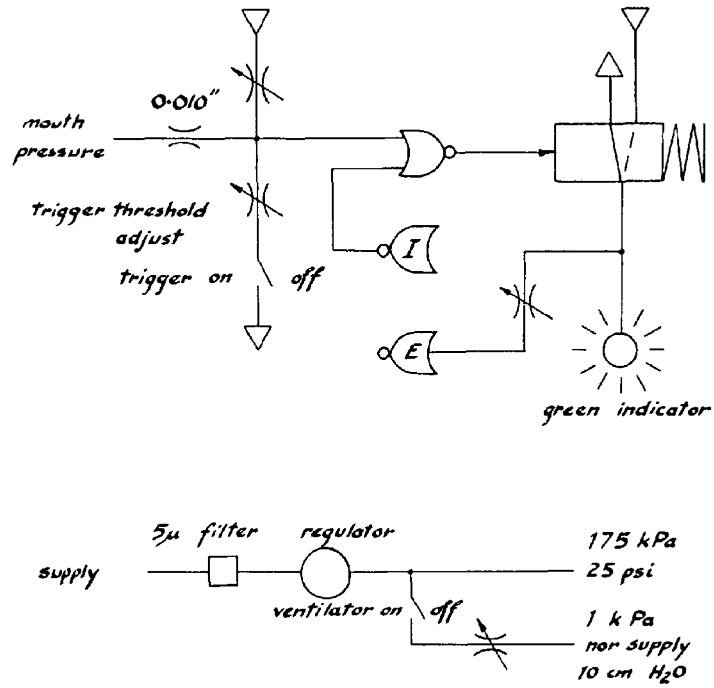


FIGURE 17. Circuit diagram for trigger and power supply for the ventilator; see text for details.

fore, at any particular flow resistance setting, flow to the patient is constant. The ventilator is thus a constant flow generator.

A push button is included in this circuit to produce start-up of the flip-flop oscillator and also to provide the means to switch to inspiration at any time.

Figure 16 shows the circuits for low and high pressure alarms. At the top of Figure 16, the circuit is designed so that unless mouth pressure develops sufficiently during inspiration, a nor gate turns on an interface valve operating a low pitched whistle. This alarm signals either patient disconnection or an oscillator failure. The lower circuit of Figure 16 details a high pressure alarm and inspiration cut-off, essential in flow-generator ventilators. The high pressure limit adjust valve (Clippard MNV-1K) determines at what mouth pressure a diaphragm controlled valve (Johnson FPC-20) will open (normally held closed by 1 kPa or 10 cm H_2O nor supply pressure at the control port) and operate an interface valve. The interface valve operates a high pitched whistle, a red indicator (Norgren fluidics) and switches the I nor off thereby cycling the ventilator to expiration.

Figure 17 shows circuits for a patient trigger (top) and the ventilator supplies to the interface valves and the nor gates (bottom). The trigger threshold adjust valve (Clippard MNV-1K) determines at what mouth pressure the inlet pressure to a nor gate will decrease enough to allow the nor gate output to switch on and operate an interface valve. The interface valve turns the E nor gate off, initiating inspiration and also operates a green indicator. The supply circuit shown at the bottom of Figure 17 accepts supply gas at pressures from 0.2 MPa to 7 MPa or 30 psi to 1000 psi and filters it with a 2–5 micron filter element (Hoke 6300 series).

Such filtering is extremely important; the reliability of fluidic-pneumatic circuits is rapidly compromised by a dirty gas supply. The filtered gas is then regulated to 175 kPa or 25 psi (Conoflow Corp. FH-60) for the interface valves supply, and the 1 kPa or 10 cm H_2O supply for the nor gates is derived from the interface valve supply by using a miniature needle valve as a pressure reducing element.

Conclusion

Fluidic and pneumatic components and circuits have matured rapidly from their beginnings, and increases in their reliability over the older mechanical-pneumatic devices have made them the rivals of electrically operated components and have led to their use in current ventilator designs. Nevertheless, they do not yet rival the complexity of electronic devices and the next generation of ventilators may be a hybrid of low power digital electronic devices and modern fluidic and pneumatic devices with the resulting benefits of increased versatility and precise control.

Summary

This article describes the construction and operation of fluidic and pneumatic devices in current use, as well as the principles of their use in medical devices. The designs for an automatic blood pressure cuff inflator, a blood pump and a high performance ventilator are presented.

Résumé

L'utilisation technologique des principes physiques de l'écoulement des fluides a contribué récemment à perfectionner les respirateurs utilisés en médecine. Il nous apparaît donc à propos d'exposer ces principes et d'en illustrer l'application dans différents appareils d'usage courant en anesthésie.

Vers 1950, l'apparition d'un appareil appelé amplificateur "fluidique" a déclenché un intérêt marqué et conduit à l'utilisation en médecine de certaines propriétés des fluides en mouvement.

Le diagramme de la figure no 1 illustre ce type d'amplificateur dont toute la conception repose sur l'effet de Coanda, du nom de l'ingénieur roumain qui l'a décrit. Un jet de fluide, (entendre par fluide un gaz ou un liquide) émanant d'un bec, a tendance à adhérer à la surface de la paroi adjacente. Ce phénomène constitue l'effet Coanda. A sa sortie, le courant fluide provoque un effet d'entraînement de gaz autour de lui et crée ainsi une pression négative sur ses bords. Cette pression négative étant plus forte du côté où la paroi est plus proche attire le jet vers cette paroi. A mesure que le courant fluide s'approche de la paroi, la pression négative diminue et l'écoulement se fait finalement au contact de la paroi.

En fin de compte, on peut concevoir l'effet Coanda comme un carollaire de l'effet venturi ou un résultat du principe de Bernoulli. Dans le cas de l'amplificateur fluidique, un jet latéral permet de contrôler la paroi vers laquelle le courant fluide ira adhérer. La figure no 1 montre qu'une impulsion momentanée émanant du jet latéral oriente le courant principal vers l'une ou l'autre des sorties. La pression minimale nécessaire dans le jet latéral justifie le terme d'amplificateur fluidique.

A noter que le terme fluidique ne devrait s'appliquer qu'à des appareils n'ayant aucune pièce mobile si ce n'est le fluide lui-même; les autres contenant des pièces mobiles sont des appareils pneumatiques.

Cette absence de parties mobiles dans l'appareil fluidique élimine l'usure et la fatigue du matériel et représente un sérieux avantage sur les appareils pneumatiques; cependant, il faut retenir que dans ces appareils le courant de fluide doit être continu, ce qui implique une dépense ininterrompue d'énergie contrairement aux appareils pneumatiques.

Après discussion de ces principes, l'auteur dans cet article illustre leur application dans trois appareils médicaux : un lecteur automatique de pression artérielle, une pompe à circulation extracorporelle et un ventilateur "fluidique".

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