

April 14, 1970

M. S. HEILMAN
INTRA-AORTIC HEART PUMP

3,505,987

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3 Sheets-Sheet 1

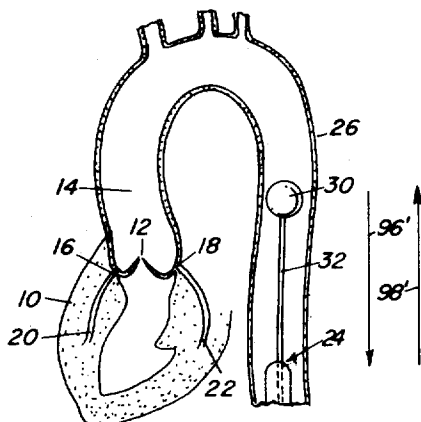


FIG. 1

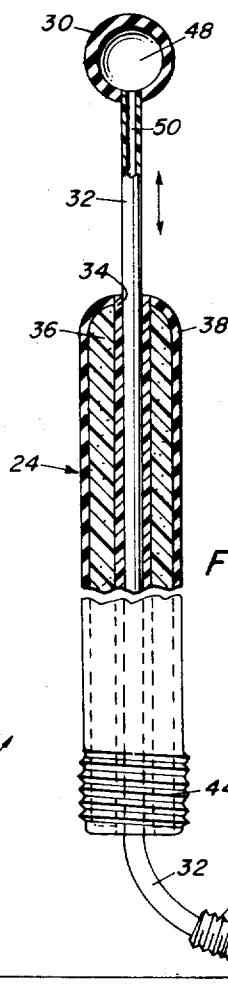


FIG. 2

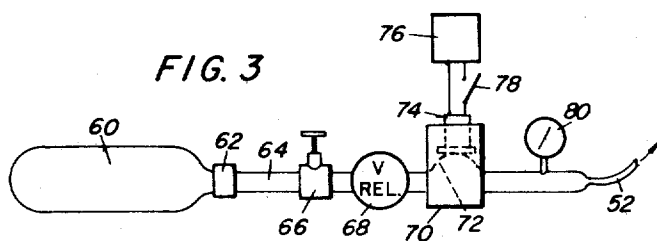


FIG. 3

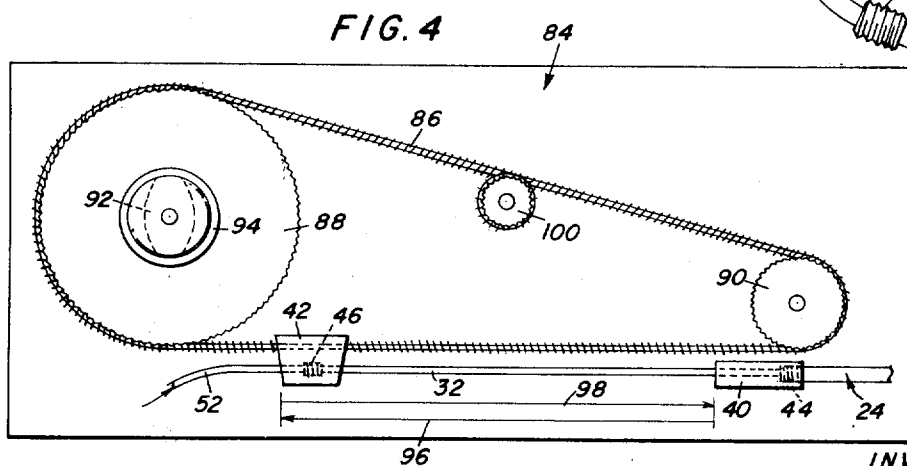


FIG. 4

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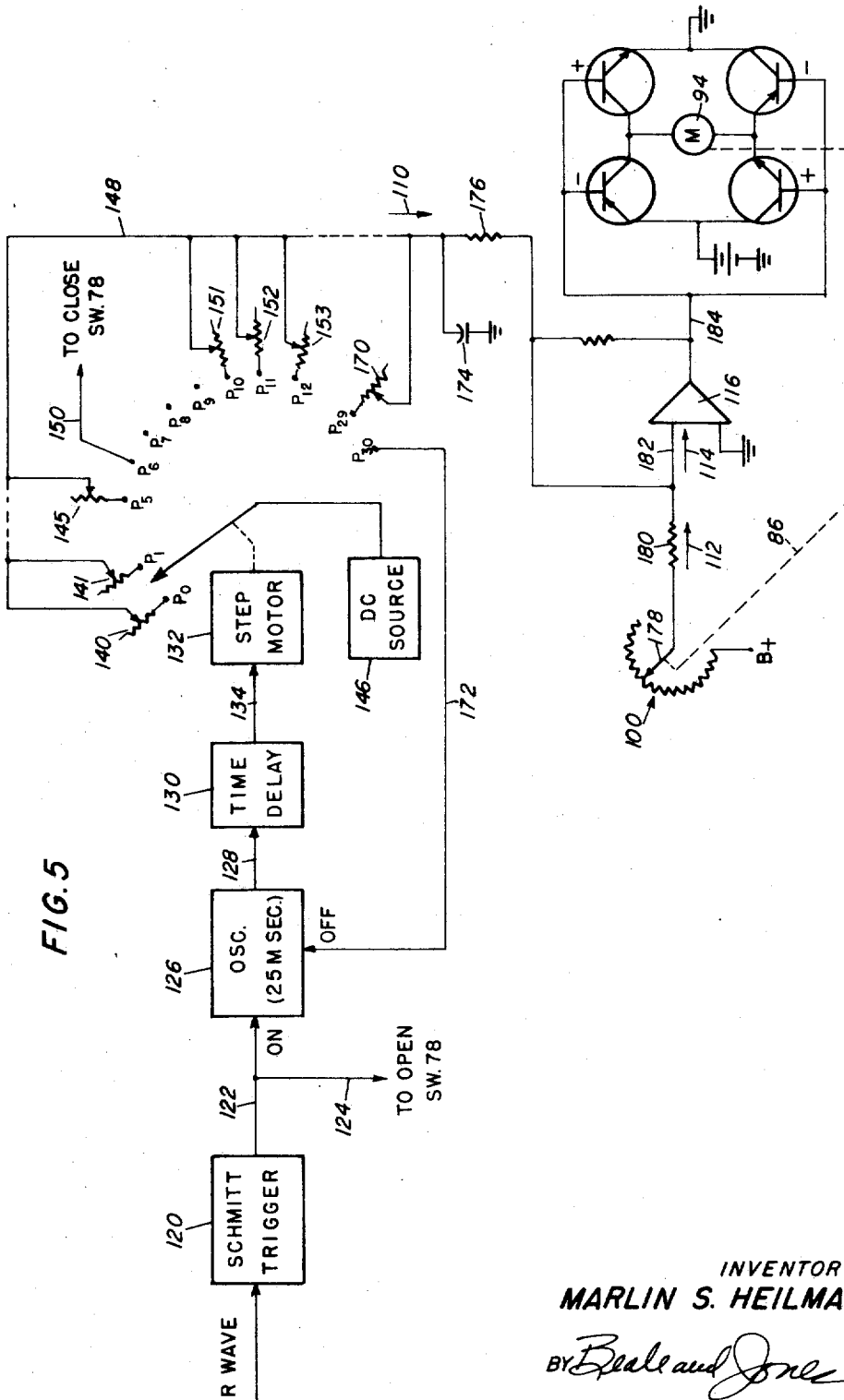
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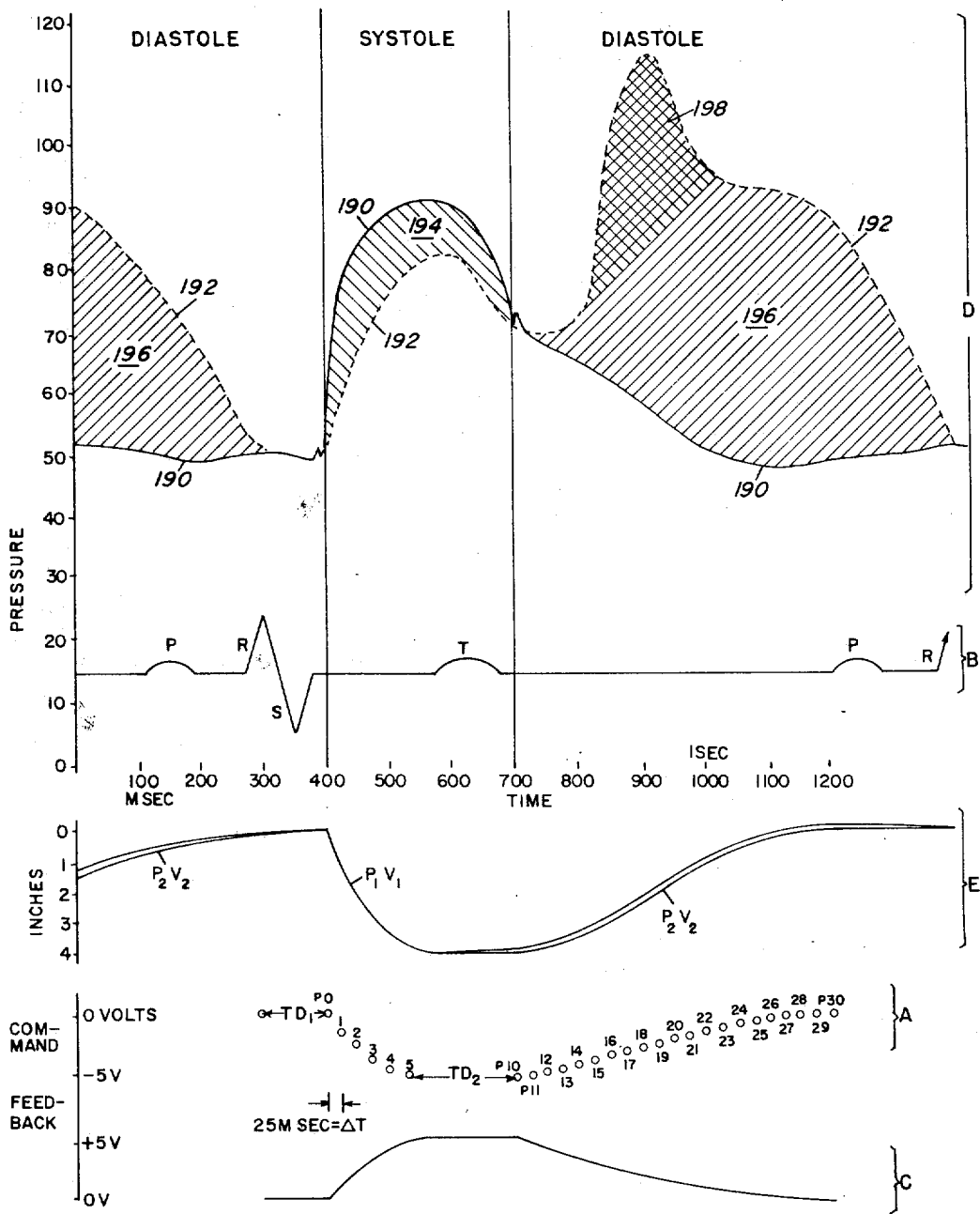


FIG. 6

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INTRA-AORTIC HEART PUMP

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35 Claims

ABSTRACT OF THE DISCLOSURE

A counterpulsation system for aiding coronary circulation wherein an expansible impeller means is located within the aorta of a patient. Pressure regulating means are provided for expanding and contracting said impeller while simultaneously reciprocating it within the aorta, the motion of said impeller being synchronized with the pumping activity of the heart by means of a variable control circuit to reduce aortic pressure during systole and increase it during diastole.

BACKGROUND OF THE INVENTION

The present invention relates to a method and apparatus for assisting the flow of blood and maintaining its pressure within the aortic artery of a cardiovascular system. More particularly, it relates to a method and apparatus for carrying out the principle of counterpulsation whereby cardiac work is aided by lowering aortic pressure during systolic ejection and by raising the diastolic pressure.

Heart disease is the leading cause of death in the United States today. A large fraction of heart deaths are caused by acute (sudden) partial or complete obstructions of blood flow to the heart muscle, causing heart muscle weakness or abnormal electrical phenomena which, in turn, make the heart nonfunctional as a pump. Many of these obstructions, hereinafter called coronary occlusions, result in death within hours or days of their occurrence. It is probable that, if the heart were assisted during the first critical hours, the reparative and regenerative powers which are inherent in the heart could preserve life in a large number of victims of such attacks.

During normal functioning of the heart, one complete heart cycle takes approximately one second and consists of two time intervals. The first interval is termed systole (approximately 300 milliseconds), during which the aortic valve is open and the left ventricle contracts and squeezes blood into the aorta. At this time, the pressure within the left ventricle is in equilibrium with the aortic pressure. The second interval (approximately 700 milliseconds) is termed diastole, and is that time following systole during which the aortic valve is closed and the left ventricle is being filled through the mitral valve. The aorta is a large extendible elastic tube that dampens the pulsatile character of the blood being ejected from the heart and provides a conduit for the high pressure arterial blood to move into the various organs of the body. Immediately above the aortic valve are located the openings of the coronary arteries which supply arterial blood to the heart, itself. It is during diastole that the greatest amount of coronary flow takes place. Not only is the time interval longer, but the intra-muscular pressure is lower since less work is involved during this period, thus allowing more blood to pass through the coronary system.

Upon the occurrence of an acute occlusion of one of the principal coronary arteries, with consequent oxygen and nutrient starvation of the respectively supplied heart muscles, the left ventricle becomes weakened and, depending on various compensatory physiological mechanisms, is often unable to eject sufficient blood to maintain

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a normal aortic pressure. Such an event has a degenerating effect on heart viability because the coronary circulation suffers further from the lower driving blood pressure head in the aorta.

It has been recognized for a number of years that the effects of this degenerating cycle could be offset through the principle of counterpulsation. This concept suggests that some means should be employed to aid systolic ejection by means of lowered aortic pressure and to increase coronary circulation by raising the diastolic pressure. Many techniques for accomplishing counterpulsation have been developed, but these have not been entirely satisfactory in that they are imprecise in their timing, do not affect both the systolic and diastolic portions of the cycle, or are ineffective. For example, some systems concentrate only on raising diastolic pressure, but such diastolic augmentation systems fail to assist in the systolic ejection of blood from the left ventricle. It is, therefore, an object of the present invention to provide an improved method and apparatus for accomplishing counterpulsation wherein both systolic and diastolic portions of the heart cycle are affected.

SUMMARY OF THE INVENTION

Briefly, the present invention provides a reciprocating pump means which can be implanted in the aorta by way of a guide tube which extends from outside the body of the patient to the area where the pumping action is to be carried out. The pump means includes a driving mechanism, a connecting tube, and a pistonlike impeller means so arranged that the intra-aortic impeller may be driven to move back and forth within the aorta. The impeller means may be an elastic balloon or bladder that is inflatable through the connecting tube so that its internal gaseous pressure determines its external dimensions. Control means are provided for the driving mechanism and a pressure regulating mechanism so that the external dimensions of the balloon may be changed in synchronism with the motion of the impeller. The pressure regulating means may be operated at the end of each stroke so that the pump means has one set of dimensions on the downstroke and a second set of dimensions on the upstroke, or it may be operated to change the impeller dimensions during its motion. Electrical programming means are provided for the driving mechanism and for the gas pressure regulating system so that the pressures generated within the aorta by the motion of the impeller means will follow a predetermined pattern, thus assisting the heart in its functional cycle and assuring a proper blood supply to the coronary arteries. Suitable means are provided to generate an electric command signal representative of the desired pressure cycle which is to be established by the position and size of the impeller means, and a second signal is generated by the driving mechanism of the pump, this second signal representing the actual position of the impeller means within the aorta. These two electric signals are compared and an error signal derived therefrom, the error signal being used to establish the position of the impeller means. By means of this feedback and comparing circuitry, the motion of the pump means is caused to follow the predetermined pattern, which pattern may be varied in accordance with the requirements of the particular patient.

BRIEF DESCRIPTION OF THE DRAWINGS

The foregoing features of the invention, as well as various modifications thereof, will be understood more clearly and fully from the following detailed description, taken in connection with the accompanying drawings, in which:

FIG. 1 is a diagrammatic illustration of a portion of

a cardiovascular system, showing the location of the intra-aortic heart pump of the present invention;

FIG. 2 is a partial cross-sectional view of the impeller means and the associated catheter device by which it is positioned in the aorta;

FIG. 3 is a diagrammatic illustration of the gas pressure regulating system for the impeller means of the heart pump;

FIG. 4 is a diagrammatic illustration of a suitable drive mechanism for the pump;

FIG. 5 is a schematic diagram of a control system suitable for use in the present invention; and

FIG. 6 is a graphical illustration of the functioning of the present invention.

DESCRIPTION OF THE PREFERRED EMBODIMENT

Referring now to FIG. 1 of the drawings, there is illustrated in diagrammatic form the muscular left ventricle 10 of a heart which, upon contraction, ejects blood through the aortic valve 12 into the aorta 14. Immediately above the aortic valve are located the openings 16 and 18 leading to the coronary arteries 20 and 22, respectively. As has been noted, the coronary arteries supply blood to the muscles of the heart itself, and it is the partial or complete blockage of blood flow through these arteries and others (not shown) which is known as a coronary occlusion.

In accordance with the present invention, the blood pressure cycle in the aorta is maintained at a predetermined cyclical level by means of an intra-aortic device, indicated generally at 24, which is located within the longitudinal portion 26 of the thoracic and abdominal aorta. As illustrated in greater detail in FIG. 2, the intra-aortic device 24 is comprised of an impeller means 30 secured to a connecting tube 32 located within, and guided by, a coaxial tube 34. Taken together, the connecting tube 32 and the guide tube 34 form a catheter device which is designed for insertion into the body of the patient, with a portion extending outside the body for connection to a suitable electro-mechanical actuator, to be described. This actuator permits the impeller means 30 to be driven back and forth within the aorta by means of the connecting tube 32 to effect the desired pressure cycle within the aorta. Impeller means 30 acts as an inflatable piston, or balloon, within the aorta, being expandable and contractable in accordance with a pressure regulating means connected to it by way of the connecting tube 32. An increase in the pressure applied by way of tube 32 to the impeller causes an increase in its external dimension, the external dimensions being proportional to and thus controllable by the internal gas pressure.

The guide tube 34 of the catheter is made of Teflon to permit easy motion of connecting tube 32 within its central opening. Surrounding the Teflon tube 34 is a fluid or elastic medium, such as liquid silicon or a polyurethane foam, which acts as a shock absorber between the Teflon tubing and an outermost enclosing tube 38. This outermost tubing is preferably a medical grade silicon rubber so that it is compatible with the blood stream. The external diameter of tubing 38 is sufficiently large that, when the connecting tube 32 is moving reciprocally during operation, any laterally transmitted force is first damped by the shock absorbing material 36 and then distributed over a large cross sectional area of aortic wall. The large circumference of the external silicon tubing and the shock absorbing material within it thus serve to prevent damage to the walls of the aorta.

Both ends of the external silicon rubber tube 38 are smoothly sealed to the Teflon guide tube 34, which, in turn, snugly sheaths the connecting tube 32. The proximal end of the catheter device 24, which end extends out of the aorta of the patient, is attached to the external driving mechanism in the manner illustrated in FIG. 4. The Teflon guide tube 34 and the external sealed silicon tub-

ing 38 are both attached to a stationary support 40 on the electro-mechanical actuator, while the movable, semi-rigid, connecting tube 32 is attached to a movable driver mechanism 42. This arrangement permits the connecting tube to be moved longitudinally inside the aorta and inside the catheter device. The catheter tubing is fastened to the stationary support 40 by any suitable means such as threads 44, while the inner connecting tube 32 is attached to the movable drive means 42 by any suitable means such as threads 46.

The expansible central lumen, or chamber 48 of the intra-aorta impeller means communicates through the axial opening 50 of connecting tube 32 and through a flexible tube 52 with a source of a biologically acceptable gas such as carbon dioxide. Such a source is illustrated diagrammatically in FIG. 3 wherein a carbon dioxide cartridge 60 is attached by means of a threaded connection 62 to a gas line 64. Line 64 includes a manually controlled needle valve 66 which is followed by a preset pressure relief valve 68. The pressure relief valve 68 is included to prevent explosion of the intra-aortic impeller should excessive pressure from the CO₂ cartridge 60 flow into the system. Line 64 includes a miniature accumulator 70 having a flexible wall 72 which is movable by means of a solenoid 74. The movable wall, or diaphragm, is in the form of a plastic bag, and is preferably of a vinyl material whereby motion of the piston may vary the volume of accumulator 70. The solenoid is operated by means of a variable direct current source 76 connected to the coil of the solenoid through an on-off switch 78. Current source 76 may be manually variable or it may be automatically varied in accordance with a predetermined program. Switch 78 may be mechanically operated or may be an electronic device such as a transistor switch; in either event, it must be of a suitable type to respond to the control signals provided by the programmed controller of FIG. 5, to be described.

A pressure gauge 80 is provided in gas line 64 downstream from accumulator 70 to monitor the pressure being applied to the central lumen of impeller 30 by way of flexible tube 52. The initial pressure and volume, hereinafter termed P_1V_1 , of the impeller is determined with switch 78 opened. The system is filled with carbon dioxide through needle valve 66 until pressure P_1 is reached, as determined by pressure gauge 80. The needle valve is then closed and the pressure regulator system is ready for operation. When switch 78 is closed, solenoid 74 produces a piston action on diaphragm 72 in accordance with the magnitude of the current supplied by source 76. This piston action increases the pressure in the system, which increase is registered on pressure gauge 80 as P_2 , pressure P_2 being proportional to the current from source 76. At pressure P_2 , the impeller means 30 will expand to a second volume V_2 and will take on a correspondingly increased external dimension. Thus, the impeller may be a bi-stable pressure-controlled device which will have two sets of dimensions depending upon whether it is subjected to pressures P_1 or P_2 and depending on its particular shape and expansion characteristics. If a programmed variable current source 76 is used, multiple pressure and volume states may be created in the system during each cycle of the heart pump operation, thus permitting a further control over the pressure pattern within the aorta. The desirability and extent of such additional variations would depend on the particular patient being treated.

Referring now to FIG. 4, there is illustrated an electro-mechanical actuator which is designed to reciprocate the pistonlike impeller means 30 within the aorta, the stroke, or distance, and the timing of the motion being controllable in accordance with a predetermined pattern or program. The actuator is mounted in a suitable housing or may be attached to a suitable platform 84, and includes the stationary connecting element 40 which is adapted to receive the threads 44 of the catheter device 24. The driver

5 means 42 is fastened to a positive drive timing belt 86 which passes around a geared drive pulley 88, an idler pulley 90 and a potentiometer 100. Pulleys 88 and 90 are so located on platform 84 that the portion of the timing belt 86 which carries driver 42 is parallel to the semi-rigid connecting tube 32 extending from stationary support 40. As has been noted, connecting tube 32 is fastened to driver 42 by means of suitable threads 46; thus rotation of drive pulley 88 causes drive means 42 to move tube 32 within the catheter device 24.

10 The geared drive pulley 88 is connected to the rotary armature 92 of a direct current, limited rotation torque motor 94. This motor, which may be a Series TQ18W brushless DC torque motor, manufactured by Aeroflex Laboratories, Inc., of Plainview, Long Island, N.Y., is a precision actuator having an angular motion limited to approximately $\pm 60^\circ$ rotation, with a linear torque output. In operation, the armature of the torque motor 94 rotates clockwise through approximately two radians in response to an input signal of one polarity, introducing a linear stroking motion to driver mechanism 42. This motion is transferred through connecting tube 32 to the intra-aortic impeller means 30, causing the impeller to move downwardly in the aorta, as viewed in FIG. 1. The direction and extent of this motion is indicated in FIG. 4 by arrow 96 and in FIG. 1 by arrow 96'. Upon reversal of the energizing signal to motor 94, armature 92 rotates in a counterclockwise direction through the same two radians, introducing a reverse linear stroking motion to the driver mechanism 42 and to the impeller 30. This reverse motion is indicated by arrows 98 and 98' in FIGS. 4 and 1, respectively.

15 The motion of armature 92, and thus of drive mechanism 42 and impeller 30, is monitored by means of potentiometer 100 which is supported on platform 84 and driven by timing belt 86. The setting of potentiometer 100 thus represents the position of the impeller within the aorta, and provides a feedback signal which can be compared with a command signal to adjust the impeller to one or more predetermined positions during the course of a timing cycle. This control system is illustrated in FIG. 5, which is seen to consist of circuitry for generating a command signal 110, circuitry for providing a feedback position signal 112, and circuit means for comparing these two signals to produce a resultant, or error, signal 114. This error signal is amplified by a suitable amplifier 116 and is used to drive the motor 94 in a direction corresponding to the signal polarity, thus causing the motor to follow a desired pattern of motion.

20 Command signal 110 is a generated function which varies in amplitude with time in conformity with a determinable pattern. Although numerous types of signal generators will be suitable for this system, a preferred method of generating the desired wave form is set forth herein. As shown, the command signal is a function generated by multiple potentiometers and time delay circuits which are sequentially gated to an output line in given time increments. The advantage of this type of arrangement is that the plurality of potentiometers permit a flexible but exact manual setting of the desired amplitude for each step of the cycle, thus permitting generation of a wave form that gives close and accurate control over the motion of impeller 30, allowing this motion to be exactly matched to the requirements of each individual patient. A complete cycle of the command signal 110 is generated in response to a single output from a Schmitt trigger 120. The Schmitt trigger produces an output pulse on line 122 in response to the standard R wave of an electrocardiogram, which wave is illustrated at curve B in FIG. 6. The output signal on line 122 may be applied through line 124 to the switch means 78 (FIG. 3) to open this switch and thus to insure that P_1V_1 conditions of pressure and volume exist in the impeller 30. In addition, the signal on line 122 activates a clock means, such as an oscillator 126, which produces a pulse every

25 milliseconds. These clock pulses are applied through line 128 to a time delay circuit 130 which is preset to the desired value and, typically, might provide a delay of 100 milliseconds, or 4 clock pulses.

30 At the end of the time delay, a gating means is activated to connect successive potentiometers to an output line to produce the desired command function. The gating means shown here, for purposes of illustration, is comprised of a step motor 132 which receives from oscillator 126 stepping pulses through line 134 at the end of the first time delay. The step motor advances a movable arm 136 to successive contacts P_0 through P_{30} of corresponding potentiometers which are so connected in the command circuit as to provide the desired wave form and control. As an example, at the end of the delay provided by delay circuit 130, movable arm 136 is connected successively to contacts P_0 through P_5 , thus connecting 6 different potentiometers 140 through 145 between a regulated source of negative voltage 146 and output line 148, each step producing on line 148 a distinctive voltage level determined by the setting of the corresponding potentiometers. When arm 136 reaches contact P_6 , a second time delay is initiated, during which time switch 78 in FIG. 3 is closed to change the pressure and volume conditions in the impeller means 30 to P_2V_2 , as has been described. This may be effected conveniently by connecting a conductor 150 between contact P_6 and the actuator means for switch 78, whereby the negative voltage on arm 136 is applied to such actuating means. It will be apparent that the output from any one of the contacts may be used for this purpose, depending upon the timing desired for operation of switch 78.

35 The time delay at this point in the cycle is obtained simply by causing arm 136 to be gated to a series of contacts P_6 through P_9 which are not connected to output line 148. It will be apparent that any desired time delay means may be used here. Similarly, it will be apparent that the time delay provided by network 130 may also be provided by a corresponding use of unconnected contacts preceding contact P_0 to which arm 136 could be gated by step motor 132. Upon connection of arm 136 to contact P_{10} and until its connection with contact P_{11} , the signal on line 148 is varied by means of potentiometers 151 through 170. When arm 136 reaches contact P_{30} , the negative voltage carried by arm 136 is applied through line 172 to oscillator 126, cutting off the oscillator deenergizing step motor 132. This ends the cycle of operation of the control network and it remains quiescent until activated by a succeeding R wave from the electrocardiogram of the patient.

40 The steps between the various voltage levels of command signal 110 may be smoothed out by means of a suitable filter network, such as that formed by a capacitor 174 and resistor 176, before being compared with feedback signal 112. The filter network may also be used to maintain the voltage on line 148 at the level set by potentiometer 145 during the time delay caused by contacts P_6 through P_9 , if desired.

45 The feedback signal from potentiometer 100 is derived by means of movable arm 178, the arm being connected through resistor 180 to the input line 182 of amplifier 184 where the signals 110 and 112 are compared. Movable arm 178 is driven by means of motor 94 through medium of timing belt 86 in the manner described above. The error signal 114 appearing on line 182 is either positive or negative, depending upon whether the electrical positive position signal derived from potentiometer 100 has a magnitude that is greater or lesser than that of the electrically negative command signal 110. The output signal from amplifier 116 appearing on line 184 is an amplified version of the input error signal 114, and is applied to the stator of motor 94 in such a manner as to cause the armature to move in a direction to reduce the error signal to zero, in the known manner.

50 Referring now to the graph of FIG. 6, curve A

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trates a typical command function which may be generated by the gated potentiometers of FIG. 5. It will be noted that the voltage produced by the successive potentiometers varies between 0 and -5 volts, depending upon the setting of the particular potentiometers. As shown in this example, the successive command voltages during systole decrease in value, causing the armature of motor 94 to rotate in a clockwise direction to move impeller means 30 downwardly in the aorta, as illustrated in curve E, until the desired displacement has been achieved. At this time, the second time delay is initiated, and the shift of the pressure state in the impeller to P_2V_2 is begun. Upon completion of the second time delay, the command function begins to drive the motor armature in a counterclockwise direction, moving the impeller upwardly in the aorta at a rate determined by the change in amplitude of the command function. The upward motion of impeller 30 is seen from curve E of FIG. 6 to occur during diastole, with the piston means expanded to the diameter governed by the P_2V_2 condition. Upon completion of the cycle at contact P_{30} , the system rests until again triggered to cycle.

The displacement distance of the intra-aortic device is measured by potentiometer 35, whose output constitutes curve C in FIG. 6. It will be seen that curve C follows the shape of command function A, although reversed in polarity, thus illustrating the manner in which the system follows the command function. In this manner the electro-mechanical actuator gives a controlled and predictable position, velocity and acceleration behaviour pattern to the intra-aortic device. This device cycles up and down within the aorta a constant distance, but its hemo-dynamic effect differs during the two parts of the cycle through variations in its external dimensions. These changes in dimension with displacement are illustrated in curve E.

The curves and areas shown in section D of FIG. 6 correspond to the pressure vs. time values that exist in the aortic root 14 (FIG. 1), during the two phases of the heart cycle, that is, during diastole and systole. Curve 190 illustrates an abnormally low aortic root pressure such as might exist in a failing heart. Curve 192 represents a new intra-aortic pressure curve that would be produced by successful counterpulsation wherein the systolic pressure is reduced and the diastolic pressure is increased. It will be noted that, during systole, there is a time-pressure integral 194 which represents the lower pressure force which the left ventricle opposes when its outlet valve 12 is open with the use of the present invention. Time-pressure integral 196, which is the area between curves 190 and 192 during diastole, represents the increased intra-aortic pressure which is available through use of the invention to drive blood into the coronary arteries 20 and 22 (FIG. 1) to overcome the effects of a coronary occlusion. By proper programming of the present system, a time-pressure integral of a pressure spike nature, such as that illustrated at 198, may be provided. Such a pressure spike can be useful in stimulating certain cardiovascular reflexes.

Although particular apparatus for carrying out the present invention has been set forth herein, it will be apparent to those skilled in the art that numerous variations may be made without departing from the spirit of the invention. Thus, for example, in place of potentiometer 100, a tachometer or accelerometer could be used with appropriate electrical circuitry to provide equivalent feedback displacement vs. time information. In place of the plurality of individually adjustable potentiometers, the command function could be automatically generated by a variety of known circuits. In addition, in certain circumstances, it may be desirable to replace the illustrated pressure-volume regulator with a second pistonlike member surrounding the impeller means 30, whereby variations in the external dimensions and consequent hemo-dynamic effect of the second member would be produced through the stroking action of piston means 30 therewithin.

It will be appreciated that the principles and apparatus

described herein may be adapted for use in influencing the flow of fluid in any body canal, and that such flow can be programmed to follow any desired pattern, although the principal purpose for which the present device is adapted is its use in assisting an acutely failing heart until the heart's own regenerative powers can eliminate the need for such an assist. Normally, the intra-aortic device of this invention would be inserted to a position relatively close to the aortic root so that the maximum effect of its operation would appear in the coronary arteries, but there may be circumstances where it would be located elsewhere. Thus, various omissions and substitutions in the device and method illustrated and described may be made without departing from the spirit of the invention.

I claim:

1. Pump means for modifying fluid flow within a human or animal body, comprising: impeller means adapted for location within a fluid flow passage; driver means for moving said impeller within said passage; control means for operating said driver means in accordance with a determinable pattern; and means for varying the dimensions of said impeller by expanding and contracting said impeller in conformity with said pattern.

2. The pump means of claim 1, wherein said driver means includes connecting means for attaching said impeller means to said driver means, said driver means being operable to impart reciprocal motion to said connecting means and said impeller means within said flow passage.

3. The pump means of claim 2, further including guide means for said connecting means, said guide means comprising a relatively stationary catheter consisting of an innermost low friction tube, intermediate shock absorbing means, and an outermost sealing tube, whereby said connecting means can move freely within said innermost tube to transmit motion from said driver means to said impeller means.

4. The pump means of claim 1, wherein said impeller means comprises a hollow, elastic member having a central lumen; wherein said means for varying the dimensions of said impeller comprises a source of pressurized gas connected to said lumen; and wherein said driver means includes mechanical actuator means connected to said lumen.

5. The pump means of claim 1, wherein said variable control means includes means for producing a command signal; means for generating a feedback signal corresponding to the position of said impeller in said passage; and means for comparing said command and feedback signals.

6. The pump means of claim 5, wherein said command signal is an electrical signal having a waveform shaped in accordance with said pattern and said feedback signal is an electrical signal having a waveform determined by the motion of said impeller.

7. The pump means of claim 2, wherein said driver means comprises an electro-mechanical actuator having a reversible DC torque motor and a positive drive timing belt connected between said motor and said connecting means.

8. A counterpulsation pump for aiding blood circulation, comprising: impeller means adapted for location within a cardiovascular system; drive means for longitudinally displacing said impeller means within said system; and control means including means responsive to the position of said impeller for moving said impeller drive means in conformity with a determinable pattern.

9. The counterpulsation pump of claim 8, wherein said drive means comprises mechanical actuator means and connector means attaching said impeller means to a movable portion of said actuator means; said control means regulating the motion of said actuator means whereby said impeller is moved in conformity with said determinable pattern.

10. The counterpulsation pump of claim 8, wherein said control means is variable to establish said pattern.

11. The counterpulsation pump of claim 8, wherein said drive means includes semi-rigid connector means between said drive means and said impeller means, whereby said impeller and said connector are moved longitudinally within said system by said drive means.

12. The counterpulsation pump of claim 8, wherein said drive means is reversible; said control means determining the direction, magnitude and velocity of motion of said drive means to move said impeller means accordingly.

13. The counterpulsation pump of claim 12, wherein said control means comprises means for generating a command signal, means for deriving a feedback signal corresponding to the position of said impeller means within said system, means for comparing said command and feedback signals to produce a resultant signal, and means for operating said drive means in accordance with said resultant signal.

14. The counterpulsation pump of claim 13, wherein said means for generating a command signal is variable, whereby any arbitrary pattern may be established.

15. The counterpulsation pump of claim 14, wherein said command signal is an electrical signal having a waveform corresponding to the blood pressure variations to be established in said cardiovascular system.

16. The counterpulsation pump of claim 14, wherein said command signal is an electrical waveform having a magnitude and polarity which is variable to produce a displacement in said impeller means which serves to reduce the pressure of blood in said cardiovascular system during systole, and which serves to increase said pressure during diastole.

17. The counterpulsation pump of claim 13, wherein said feedback signal is an electrical signal having a waveform corresponding to the motion of said impeller means.

18. The counterpulsation pump of claim 13, wherein said control means further includes means for varying the dimensions of said impeller means during the motion of said impeller means within said system.

19. The counterpulsation pump of claim 18, wherein said impeller means consists of a hollow, elastic piston-like member having a central lumen, said means for varying the dimensions of said impeller means including a source of pressurized gas, tube means connecting said source of gas with said central lumen, and means for regulating the pressure of said gas, whereby the size of said impeller corresponds to the pressure of the gas within said central lumen.

20. The counterpulsation pump of claim 18, wherein said drive means for displacing said impeller means comprises electro-mechanical actuator means external of said cardiovascular system; semi-rigid connector means attaching said impeller means to a movable portion of said actuator means; and catheter means adapted to extend from a stationary portion of said actuator means into said cardiovascular system to guide said connector means.

21. The counterpulsation pump of claim 20, wherein said impeller means consists of a hollow, elastic, piston-like member having a central lumen, said semi-rigid connector means comprising a tube having a passage communicating with said central lumen, whereby gas under pressure may be fed to said central lumen to inflate said impeller means.

22. The counterpulsation pump of claim 20, wherein the movable portion of said electro-mechanical actuator means includes a reversible motor, having limited rotation, said motor being attached to said semi-rigid connector means for reciprocally driving said impeller means.

23. The counterpulsation pump of claim 22, wherein said means for deriving a feedback signal comprises potentiometer means having a movable arm driven by said reversible motor, whereby the position of said arm corresponds to the location of said impeller means.

24. The counterpulsation pump of claim 20, wherein said catheter means comprises an innermost low friction

tube, intermediate shock absorbing means, and an outermost sealing tube, whereby said connector means can move freely within said innermost tube to transmit motion from said drive means to said impeller means without damage to said cardiovascular system.

25. The counterpulsation pump of claim 13, wherein said means for generating a command signal includes an input line, an output line, a plurality of potentiometers, and gating means for connecting successive potentiometers between said input and output lines, said output line connecting said command signal to said means for comparing.

26. The counterpulsation pump of claim 25, wherein said gating means includes means for generating clock pulses; step motor means responsive to said clock pulses to connect said potentiometers; and means for initiating and means for terminating the generation of said clock pulses.

27. The method of effecting counterpulsation in a cardiovascular system, comprising:

(a) inserting a reciprocable impeller means within the aortic artery of said system;

(b) deriving a first electric signal corresponding to a predetermined point in the diastole-systole cycle of said system;

(c) generating in response to said first electric signal a second, varying electric signal corresponding to a predetermined intra-aortic pressure cycle;

(d) generating a third, varying electrical signal corresponding to the position of said impeller means within said system;

(e) comparing said second and third signals to produce an error signal;

(f) and moving said impeller means in a direction to reduce said error signal, whereby the motion of said impeller means is in conformity with said predetermined intra-aortic pressure cycle.

28. The method of claim 27, further including the step of expanding and contracting said impeller means in timed relationship to the reciprocating motion of said impeller means.

29. The counterpulsation pump of claim 8, wherein said impeller means consists of a hollow, elastic pistonlike member having a central lumen, said drive means for longitudinally displacing said impeller means comprising a semi-rigid connector, said pump further including catheter means adapted to extend into said cardiovascular system to guide said connector means.

30. The counterpulsation pump of claim 29, wherein said catheter means comprises an innermost low friction tube, intermediate shock absorbing means, and an outermost sealing tube, whereby said connector means can move freely within said innermost tube to transmit motion from said drive means to said impeller means without damage to said cardiovascular system.

31. The counterpulsation pump of claim 30, wherein said semi-rigid connector means has a passage communicating with said central lumen, whereby gas under pressure may be fed to said central lumen to vary the dimensions of said impeller means.

32. An intra-aortic pump for counterpulsation of a heart, comprising a hollow, elastic, pistonlike impeller having a central lumen, a drive means for longitudinally displacing said impeller, semi-rigid connector means attaching said impeller to said drive means, and catheter means adapted to extend into a cardiovascular system to guide said connector means.

33. The pump of claim 32, wherein said connector means includes a passage communicating with said lumen, said pump further including a source of pressurized gas and means for regulating the pressure of said gas, whereby the dimensions of said impeller may be varied.

34. The pump of claim 32, wherein said catheter means comprises an innermost low function tube, intermediate

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shock absorbing means, and an outermost sealing tube, whereby said connector means can be moved freely within said innermost tube for longitudinal displacement of said impeller to reduce systolic pressure and to increase diastolic pressure, said shock absorbing means serving to prevent damage to said cardiovascular system due to the motion of said connector means.

35. The pump of claim 34, wherein said drive means comprises an electro-mechanical actuator adapted to reciprocate said impeller, said pump further including means for varying the external dimensions of said impeller in synchronism with the longitudinal displacement of said impeller.

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