(54) Title: CIRCULATORY ASSIST METHOD AND APPARATUS
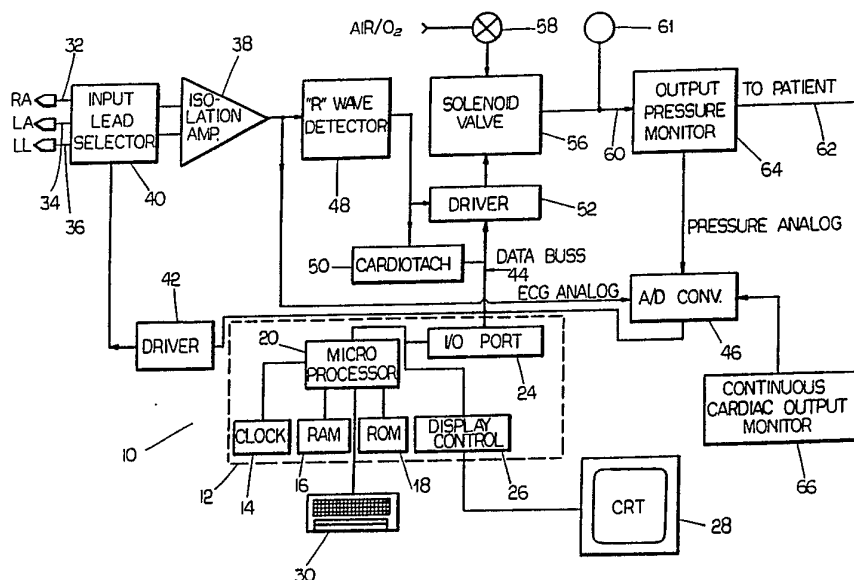
(57) Abstract

Improved cardiocirculatory assistance is provided to a patient by detecting the onset of ventricular ejection in the cardiac cycle of the patient and selectively increasing intrathoracic pressure of the patient in relative phase with respect to the onset of ventricular ejection. Cardiocirculatory output may be observed by a cardiac output monitor (66), while the relative phase of the increase in intrathoracic pressure with respect to the onset of ventricular ejection is adjusted for maximum cardiac output under the control of microcomputer (12). Output pressure line (62) is used to interface system (10) to the patient. The patient interface may include apparatus for supplying high frequency respiration pulses in synchronism with the cardiac cycle. In one embodiment, the pulses are used to inflate a bladder in contact with the thorax.

CIRCULATORY ASSIST METHOD AND APPARATUS

Technical Field

This invention relates to apparatus for enhancing
circulation in patients having a low level of cardiac out-
put. More particularly, this invention relates to an appa-
ratus and method for enhancing cardiac output by controlling
intrathoracic pressure in a manner related to the cardiac
cycle of the patient.

Background Art

Mechanical inspiration (positive pressure lung
inflation), conventionally applied to patients with acute
cardio-respiratory failure, distends the lungs, increases
airway pressure, and produces an increase in mean intra-
thoracic pressure. However, the benefit derived by the
patient from this increase in functional lung volume is
actually offset by the increase in intrathoracic pressure,
since the latter produces a decrease in venous return to the
heart, a decrease in transmural right atrial pressure (right
ventricular filling pressure) and limits right ventricular
stroke output. Because the pulmonary circulation serves as
a conduit from the right ventricle to the left ventricle and
has a high capacitance relative to the right heart, trans-
mural left atrial pressure (left ventricular filling pres-
sure) and left ventricular stroke output are similarly de-
creased. There is a 2-5 beat phasic delay of this effect
relative to the right heart. This is the primary mechanism
of decreased cardiac output in all patients during mechani-
cal ventilation in which positive pressure lung inflation is
applied either continuously (CPPV) or intermittently (IMV).

In general, cardiac output may be reduced by 50-

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75% during mechanical ventilation, which may have profound
effects on organ blood flow, tissue perfusion, and patient
survival. Further, declines in cardiac output will be ac-
centuated in settings of functional hypovolemia, myocardial
ischemia and infarction, decreased vasomotor tone, large
tidal volume breathing, prolonged inspiratory time, and the
application of positive end-expiratory pressure (ubiquitous-
ly applied to patients with acute respiratory failure to
maintain oxygenation and pulmonary gas exchange).

Changes in aortic pulse pressure and mean arterial
blood pressure can also occur during positive pressure in-
spiration. For example, peak positive pressure inspiration
produces a net increase in the series resistance of the
pulmonary circulation. Because the pulmonary circulation
mechanically couples right and left heart output, an in-
crease in pulmonary vascular resistance, such as that which
occurs with conventional low rate, high volume mechanical
ventilation (i.e., CPPV, IMV), will impede right ventricular
ejection and increase right ventricular dimension and wall
stress to produce myocardial ischemia. Decreased right
ventricular ejection will decelerate pulmonary blood flow
and decrease the filling of the left ventricle, establishing
a condition of right and left ventricular interference.
This results in decreased left heart preload (decreased left
ventricular filling due to left ventricular volume encroach-
ment by the overdistended right ventricle) and decreased
left heart ejection. In this way, conventional mechanical
ventilation may produce profound cardiocirculatory dysfunc-
tion by altering both the series and parallel relationships
of right and left ventricular function.

There have been some attempts to alleviate the
adverse effects of mechanical inspiration on circulation by
momentarily decreasing the pressure of the respiration gas
with the aid of a pulse-synchronized signal during the pro-
pagation time of the pulse wave in the alveolar flow path,

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and then immediately re-establishing the pressure that exis-
ted prior to the decrease. Proper synchronization of pulses
assists in filling of the left atrium of the heart. This is
somewhat beneficial, but does not deal with the overall
problem of decreased biventricular preload secondary to
increased mean intrathoracic pressure.

Disclosure of the Invention

It is a principal object of the invention to use
momentary timed increases in intrathoracic pressure to in-
crease cardiac output.

It is another object of the invention to utilize a
mechanical respirator to assist, rather than to depress,
cardiac circulation.

It is a further object of the invention to provide
pressure impulses to increase intrathoracic pressure at
times during the cardiac cycle which will assist, rather
than impede, cardiac filling and ejection.

In accordance with the method of the invention,
improved cardiocirculatory assistance is provided to a pa-
tient by detecting the onset of ventricular ejection in the
cardiac cycle of the patient and selectively increasing
intrathoracic pressure of the patient in relative phase with
respect to the onset of ventricular ejection. The method
may further comprise the steps of monitoring cardiocircula-
tory output and adjusting the relative phase of the increase
in intrathoracic pressure with respect to the onset of ven-
tricular ejection to maximize cardiac output. The step of
selectively increasing the intrathoracic pressure is per-
formed once every N cardiac cycles where N is a positive
integer.

In accordance with the invention, an apparatus for
improving the cardiac output of the patient comprises detec-
tion means for detecting the onset of ventricular ejection
in the cardiac cycle of the patient, pressure increasing

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means for selectively increasing the intrathoracic pressure
of the patient, and control means responsive to the detec-
tion means for activating the pressure increasing means in a
relative phase with respect to the ventricular ejection.
The apparatus may further comprise cardiocirculatory moni-
toring means for ascertaining the extent to which the appa-
ratus improves cardiac output and for providing an output
signal indicative thereof. The apparatus may also include
delay means responsive to the output signal for delaying
activation of the control means in response to the detection
means so that the cardiac output is maximized. The appara-
tus may also comprise divider means responsive to the de-
tection means for providing an output every N cardiac cy-
cles, in which case the control means is responsive to the
divided output. Thus, according to the invention, the
intrathoracic pressure of the patient is increased syn-
chronously with ventricular ejection i.e., during isovolu-
metric contraction (the time between closing of the mitral
valve and opening of the aortic valve). This may be accom-
plished by a mechanical respirator, which also assists in
respiration, or by other means such as a bladder surrounding
the thorax of the patient or an implant within the thorax.

If mechanical inspiration is shortened to 250-350
msec (so that the entire respiratory cycle occurs within the
cardiac cycle) and sychronized to the specific point in the
cardiac cycle which marks the onset of ventricular ejection
(40-100 msec post-R wave of a continuously recorded ECG),
then lung inflation and increases in mean intrathoracic
pressure occurring in phase with the cardiac cycle can be
used to maintain left ventricular preload and thereby pro-
vide a mode of augmented cardiac output. Further, because
increases in mean intrathoracic pressure occur during right
and left ejection, the transmural pressure gradient (peak
ventricular systolic pressure - aortic pressure) against
which the myocardium must work is decreased. Thus, ventri-

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cular afterload may be reduced and this may result in de-
creased myocardial wall stress and decreased myocardial
oxygen consumption. This mode of ventilation may provide a
myocardial sparing effect. Because intrathoracic pressure
is at a nadir during the diastolic phase of each cardiac
cycle, ventricular preload (filling) is augmented on a beat-
to-beat basis. The net effect is increased cardiac output
and decreased myocardial work.

Brief Description of Drawings

        Further objects, features and advantages of the
invention will become apparent upon consideration of the
following detailed description in conjunction with the draw-
ings, in which:
        FIG. 1 is a block diagram of an apparatus in ac-
cordance with the invention;
        FIG. 2 illustrates a bladder useful for increasing
intrathoracic pressure, placed around a patient, the bladder
being for use with the system of FIG. 1;
        FIG. 2A illustrates components of an embodiment of
the continuous cardiac output monitor of FIG. 1;
        FIG. 3 illustrates the manner in which the appara-
tus of FIG. 2 and FIG. 2A may be connected to the system of
FIG. 1;
        FIGS. 4A to 4E illustrate the interrelationship of
physiological parameters when a first mode of operation of
the system of FIG. 1 is commenced;
        FIGS. 5A TO 5E are similar to FIGS. 4A to 4E but
illustrate a change from the first mode of operation to a
second mode of operation of the system of FIG. 1;
        FIG. 6A and FIG. 6B are a flow chart of the pro-
gram used by the microprocessor of the system of FIG. 1;
        FIG. 7 is a flow chart of a first subroutine used
in the flow chart of FIG. 6; and

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FIG. 8 is a flow chart of a second subroutine used in the flow chart of FIG. 6.

Detailed Description of the Drawings

Referring to FIG. 1, the system 10 acᶜording to the invention provides short duration high frequency air pressure pulses which are synchronized to the cardiac cycle of the patient. The system is controlled by a microcomputer 12. Microcomputer 12 includes a clock 14, a RAM 16, and a ROM 18 connected to microprocessor 20. Microprocessor 20 is interfaced to an internal data bus 22 which is also connected an input/output port 24. Internal bus 22 is also connected to a display controller 26 which controls a display unit 28 which may be a CRT or other suitable display. As described below with respect to FIGS. 4A to 4E and FIGS. 5A to 5E, waveforms appropriate for enabling a physician or other health care professional to monitor and optimize the operation of the system may be displayed on display unit 28. Further, data and other information entered by means of keyboard 30, and interfaced to microprocessor 20 of microcomputer 12, may also be displayed. Although not shown in FIG. 1, microcomputer 12 may be of the PC type and may have a disk drive (not shown) for additional data storage capability. The disk drive may be of the type that uses floppy disks, thus providing archiving ability. Further, microcomputer 12 may have an interface for a printer.

Electrical signals indicative of a patient's heartbeat are provided to system 10 by electrocardiograph leads. These leads include RA lead 32, LA lead 34 and LL lead 36. Two of these three leads are connected to an isolation amplifier 38 by an input lead selector 40. Isolation amplifier 38 may be any of several well known amplifiers suitable for electrocardiography. Input selector 40 consists of a switching apparatus controlled by logic signals from a driver 42. Driver 42 is connected to an external

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data bus 44 from port 24. Thus, appropriate entries at keyboard 30, as described below with reference to FIG. 6, may be used to select which two of the three leads 32, 34 and 36 are connected at any particular point in time, to the input of isolation amplifier 38.

The output of isolation amplifier 38 is supplied to an analog-to-digital converter 46. Converter 46, which has an analog multiplexer input, is used to convert more than one analog signal to digital form. Converter 46 has a sufficiently high conversion rate so that all of the inputs can be converted, thus enabling the input signals to be accurately represented in digital form. The digital output of converter 46 is supplied to appropriate input terminals of port 24 by data bus 44. Microprocessor 20 then serves the function of storing the digital signals in RAM 16. These signals are eventually supplied to display controller 26 by internal data bus 22 for use in providing a display on display unit 28.

The output of isolation amplifier 38 is also supplied to R-wave detector 48. R-wave detector 48 includes a conventional peak detector which recognizes a reversal in the sign of the derivative or slope of the signal supplied by isolation amplifier 38 and provides a short digital output pulse when such reversal takes place. Prior to evaluating the wave for such reversal of sign, a filter internal to R-wave detector 48 is used to filter out the P and T waves so that only the frequency components of the QRS complex are passed to the peak detector.

The output of R-wave detector 48 is supplied to a cardiotach circuit 50. Cardiotach circuit 50 provides a digital output on data bus 44 representative of the heart rate of the patient. This may be accomplished by gating the output of R-wave detector 48 into an up/down counter within cardiotach circuit 50 for a predetermined period of time to increase the count, and then gating pulses from a clock of

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known frequency into the same up/down counter for a prede-
termined period of time to decrease the count to zero. Such
cardiotach circuits are well known in the art.

The output of R-wave detector 48 is also supplied
to a solenoid driver circuit 52. Driver circuit 52 operates
a solenoid of solenoid controlled valve 56 to open momen-
tarily to permit a quantity of gas from the output of a
pressure regulator 58 to pass through valve 56. The high
pressure input to regulator 58 is connected to an air or air
and oxygen mixture source (not shown) as may be appropriate,
depending upon the manner in which system 10 is interfaced
to the patient.

Driver circuit 52 responds to signals from micro-
computer 12 provided on data bus 44. However, as noted
above, the output of R-wave detector 48 is also provided to
circuit 52. A pulse from R-wave detector 48 does not, by
itself, activate circuit 52. Instead, a predetermined delay
interval is started, during which interval, if an appro-
priate signal is received by circuit 52 by way of data bus
44, driver circuit 52 will activate solenoid 54. Appro-
priate logic circuitry for performing this function is well
known in the art.

The output of valve 56 is connected to a pressure
line 60. A pressure reservoir or ballast 62 is also con-
nected to line 60 to provide some smoothing of the air pul-
ses from valve 56, thus providing more comfort for the
patient.

Line 60 is connected to an output pressure monitor
64. Monitor 64 includes an appropriate transducer for pro-
viding an analog signal representative of the gauge pressure
in an output pressure line 62 which is used to interface
system 10 to the patient. The analog output of the pressure
transducer within monitor 64 is provided to converter 46 for
purposes of being converted into a digital format which can
be processed by microcomputer 12. Thus, if appropriate data

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are stored in RAM 16, the analog pressure waveform can be
displayed on display unit 28. Further, alarm thresholds can
be set to notify the medical staff if a condition that is
detrimental to the patient's well-being occurs or to ini-
tiate automatic responses to alleviate or correct any such
occurrence under the control of a program stored in ROM 18.

A continuous cardiac output monitor 66 (FIG. 3)
provides an analog signal to one input of analog-to-digital
converter 46. As is the case for pressure monitor 64, out-
put monitor 66 may be used to provide an alarm function or
to initiate predetermined changes in the operation of system
10 if output signals from output monitor 66 are such as to
indicate that conditions detrimental to the health of the
patient exist.

Referring to FIG. 2, output line 62 is connected
to an inflatable pressure jacket or bladder 68 which is
secured around the thorax of a patient 70 by suitable re-
lease straps 72. The use of an inflatable bladder to inter-
face the patient to system 10 is extremely advantageous in
that it is completely noninvasive. Bladder 68 can be any of
several well known bladders such as those used to treat
shock or those used by the pilots of high performance air-
craft to control the distribution of blood in the body under
conditions of high acceleration. By the proper application
of pressure pulses along output line 62, intrathoracic pres-
sure is rhythmically increased at timed intervals related to
the cardiac cycle, as more fully described below.

FIG. 2A illustrates an alternative approach for
interfacing system 10 to the patient. This embodiment
includes a continuous cardiac output monitor 66. A first
oxygen concentration sensor 74 measures the concentration of
oxygen in the air/oxygen mixture of line 62. A patient
attachment tube 76 which opens into line 62 is in turn con-
nected to an endotracheal tube (not shown) which is used to
provide mechanically assisted respiration to the patient. A

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second oxygen sensor 78 in line 62 determines the concentration of oxygen in the air stream exhausted by tube 62, which is a combination of the air stream from pressure monitor 64 (FIG. 1) and the gas exhausted by the patient's lungs. A flow sensor 80 determines the volume of gas exhausted from tube 62. Flow sensor 80 also serves to provide a small amount of resistance to flow so that when pressure pulses are provided due to the opening of valve 56 (FIG. 1) a portion of the gases of the pulse are forced into the patient's lungs through tube 76 and the endotracheal tube.

Oxygen sensors 74 and 78 may be those associated with a polarographic oxygen analysis system or by techniques of optical fluorescence either of which may provide the actual inputs to microcomputer 12. Flow sensor 80 may be any of several well known devices such as a pneumotach device and is associated with an appropriate electronics package to provide an input to microcomputer 12.

The continuous cardiac output monitor 66 of FIG. 2A also includes a means for determining the difference in arterial and venous blood oxygen concentration. A finger tip pulsed oximeter sensor 82 (with an associated electronics package, not shown) is used to determine arterial blood oxygen content. An optical oxygen concentration sensor 84, positioned in the pulmonary artery, at the end of a pulmonary artery catheter 86 (and an associated electronics package, not shown) is used to determine venous oxygen content.

Signals from first oxygen sensor 74, second oxygen sensor 78, flow sensor 80 and the electronic packages associated with sensors 82 and 84 are all provided to microcomputer 12. Using a variation of Fick's equation, microcomputer 12 monitors, on an almost instantaneous basis, the rate of oxygen uptake and the difference in arterial and venous oxygen concentration to determine cardiac output.

Fick's equation is based on the theory that the

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uptake or release of a substance by an organ is the product
of the blood flow to the organ and the arteriovenous concen-
tration of the substance. This principle, when applied to
the lungs, can be stated as: oxygen uptake is equal to the
product of blood flow multiplied by the difference in arter-
ial oxygen concentration and venous oxygen concentration.
For purposes of cardiac output determinations, the cardiac
output is equal to the oxygen consumption of the patient
divided by the arterial-venous oxygen concentration differ-
ence.

Monitoring is accomplished by multiplying the
difference in oxygen concentration at sensor 74 and sensor
78 by the flow rate output of sensor 80 and dividing this
product by the difference in arterial and venous oxygen
concentration as determined by sensors 82 and 84. Succes-
sive quotients are stored in a portion of RAM 16 programmed
to operate as a first in/first out memory. When the sum of
the values stored falls below a predetermined limit, an
alarm is provided, thus indicating that the patient's car-
diac output has dropped.

While the continuous cardiac output monitor 66 of
FIG. 2A is designed to provide a relatively accurate abso-
lute determination of cardiac output, other systems may be
used in place of the apparatus illustrated in FIG. 2A.
Relative indications of cardiac output may be provided to
microcomputer 12 by an in dwelling optical flow sensor, a
Doppler flow probe, an electromagnetic flow meter or an
impedance flow sensor. However, the electronic systems
associated with these sensors are responsive to flow rate,
rather than to cardiac volume, and the outputs (typically
averaged over a period of time) provide a relative, rather
than an absolute, representation of cardiac output.

The respiration induced by the patient interface
in accordance with FIG. 2A consists of short rapid breaths.

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This is a marked departure from the operation of convention-
al respirators wherein the natural breathing cycle of the
patient is utilized. The short rapid breaths, if synchro-
nized to cardiac ejection as described below, provide an
extremely significant enhancement of cardiac output.

While FIG. 2 and FIG. 2A illustrate, respectively,
one completely noninvasive patient interface and one patient
respiratory interface which is relatively noninvasive, it
will be recognized that other means for increasing intra-
thoracic pressure are contemplated. For example, an appro-
priate bladder may be implanted within the thorax of the
patient in proximity to the heart so as to increase intra-
thoracic pressure in accordance with the principles of the
invention. However, it will be recognized that this
approach entails significant risks due to the necessary
surgical procedures.

While the manner of interfacing to a patient de-
scribed in FIG. 2 or FIG. 2A can each be implemented inde-
pendently, it is also contemplated that both may be used
simultaneously to increase intrathoracic pressure in unison.
Referring to FIG. 3, output line 62 of system 10 may be
connected to both bladder 68 (FIG. 2) and to a patient at-
tachment tube 76 (FIG. 3) using for example, a simple T
connector 88 in line 62. Suitable adjustable flow constric-
tors 90 may be provided downstream of T connector 88 to
properly apportion the flow of gas in line 62 between blad-
der 68 and patient connection tube 76.

The arrangement of FIG. 2A may be used indepen-
dently as a cardiac output monitor when connected to a con-
ventional respirator. In this regard, even if the only
patient interface used for circulatory assist in accordance
with the invention is the bladder of FIG. 2, the continuous
cardiac output monitor of FIG. 2A is advantageously utilized
to monitor the operation of the system, as more fully de-
scribed below with reference to FIG. 6.

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        Referring to FIGS. 4A to 4E, the interrelationship of
physiological parameters and the operation of the system of
FIG. 1 is illustrated. From top to bottom, the traces pre-
sented represent the aortic pressure A_O, the electrocardio-
gram ECG, the square wave SQ activating the solenoid of
valve 56, the ventilation breath Q_v and the cardiac output
Q_F as measured by an electromagnetic flow probe. The opera-
tion of system 10 commences at time T. Before this time
aortic pressure A_O is low, indicating an insufficiency in
cardiac output such as may be present in a patient with
circulatory disease. The occurrence of a first QRS complex
in the ECG following time T triggers the beginning of a time
delay interval t_d. At the end of this delay interval (which
can be anywhere in the range of from about 40 ms to about
200 ms, but is typically on the order of 100 ms) a square
wave is provided to the solenoid of valve 56, thus opening
valve 56. Valve 56 remains open for a period of time t_{ip}
(an inspiratory plateau time of typically 120 ms) determined
by microcomputer 12. For every QRS complex in the ECG after
time T, another square wave delayed by time t_d and of width
t_{ip} is provided. The resulting pressure pulses in line 62
cause the patient's respiration to change from that of the
normal respiratory cycle prior to time T to that of short,
high frequency breaths occurring at a rate equal to that of
the cardiac rate after time T. After such high frequency
respiration commences (delayed in time with respect to ven-
tricular ejection by a time corresponding to interval t_d) a
substantial increase in cardiac output Q_F is observed.
        The operation of the system, as described above,
is in a 1:1 respiration to cardiac rate mode. As illustra-
ted in FIGS. 5A to 5E, this is not the only mode of opera-
tion. System 10, by suitable programming of microcomputer
12 (which is either manually selectable through information
provided at keyboard 30, or may be automatically deter-
mined), can operate in a mode wherein one respiration pulse

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is provided for every N cardiac cycles. Typically N may be
equal to 2. As represented in FIGS. 4A to 4E, the transi-
tion from 1:1 to 1:2 operation occurs at time U. In the 1:2
mode there is a solenoid square wave SQ provided for every
other QRS complex of the ECG waveform. The respiration Q_v
follows the solenoid square wave SQ. It will be understood
that the depth of respiration required for supplying an
adequate amount of oxygen will be larger for the 1:2 mode
than for the 1:1 mode, and therefore a longer pulse width is
appropriate. However, this consideration applies only if
the embodiments of FIG. 2A or of FIG. 3 are used as the
patient interface. If, on the other hand, the patient is
breathing on his own or a separate respirator is provided,
and a patient interface in accordance with FIG. 2 is used,
no adjustment in the pulse width is required.

The use of a 1:2 mode is most appropriate when the
patient has a rather rapid heartbeat. Due to inertia asso-
ciated with the bulk transport of gas in the lungs, it is
preferable that system 10 operates at an output pulse rate
of approximately 40-60 per minute. Thus, if the patient has
a heart rate greater than approximately 85 bpm, operation in
the 1:2 mode may be appropriate. As illustrated in FIG. 5E,
the cardiac output Q_F increases even further when the system
is switched from the 1:1 mode to the 1:2 mode.

Referring to FIGS. 6A and 6B a program for execu-
tion by microprocessor 20 of microcomputer 12 may be loaded
from disk (not shown) into RAM 16 or may be permanently
stored in ROM 18. At step 100, when the program is called
up by using an appropriate entry on keyboard 30, the program
is started by initializing all registers. An initial screen
display is provided on display unit 28 so that data concern-
ing the patient can be input to the system using keyboard 30
at step 102. Appropriate fields for the patient's name and
relevant data, (e.g. the date, the time, ECG lead selection
etc.) are provided. After the last entry has been provided

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or, by default, if no data are provided, a switchover to a physiologic display occurs at step 104. At this point, if the electrocardiograph leads have been connected to the patient, the physician may view the patient's electrocardiogram on display 28. To assist the physician, the display may include a digital read-out of the patient's pulse rate.

At step 106 a loop (not separately shown) for monitoring the output of the continuous cardiac output monitor 66, which should have been connected to the patient, is started. As described above with respect to FIG. 2A, monitoring is accomplished by adding the entries in a first in/first out memory to determine the cardiac output of the patient. This loop operates continuously from this point forward with the current value being accessed from time to time as described below.

Microcomputer 12 may be programmed so that if the continuous cardiac output monitor 66 has not been connected, only continuous manual operation is permitted.

At step 108 a control mode and physician-selected control data may be entered. A decision may be made at this point to utilize manual control. In this case, the physician will enter a value for N of, for example, one or two to select the 1:1 or 1:2 mode respectively as described above. Further, selected values for t_d and t_{ip}, in milliseconds, may be entered. Default values may be provided which are automatically entered if the data are not changed. After this data, which are placed in appropriate fields on the display of display unit 28, is checked, it is entered. The current value of cardiocirculatory output Q_n, as determined by continuous cardiac output monitor 66, is stored and system 10 starts to provide cardiocirculatory assistance at step 110. There is a delay at step 112 of approximately fifteen seconds to allow the system to operate and bring the patient to a new state of equilibrium. At step 114 a check is made to determine whether the output of continuous car-

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diac output monitor 66 (provided continuously by the loop
initiated at step 106) indicates that the cardiac output of
the patient is less than or equal to that just prior to the
start of the assist at step 110. If for some reason the
cardiac output has not increased, the assist is terminated
at step 116 and an alarm is activated in the form of an
audio tone emitted by display unit 28 as well as an appro-
priate visual alarm indication thereon (step 118). Thus, if
the operation of system 10 does not provide increased car-
diac output, the system operator is immediately alerted. It
is then necessary either to restart the system and select
different control data, or to check for the possibility of a
system malfunction.

        Assuming the assisted cardiac output is greater
than the unassisted cardiac output, the program branches to
step 120 where a determination is made as to whether the
manual or automatic mode was selected. Assuming that the
manual mode was selected, the program branches to step 121
where the output of the continuous cardiac output monitor 66
(as determined by the loop continuously running from the
time of execution of step 106) is continuously monitored.
As long as cardiac output is greater than 0.7 times the
assisted cardiac output determined after step 114, operation
of the system continues without any alarm being produced.
This allows for some decrease in the assisted cardiac output
due to, for example, the patient falling asleep. However,
if cardiac output does decrease below 0.7 times the assisted
value, the program branches to step 118 and the alarm is
sounded. In this case, cardiocirculatory assistance is not
automatically terminated.

        Although not illustrated in FIGS. 6A and 6B, it is
advantageous to provide a program interrupt at some point,
such as during monitoring by step 122 so that the input data
can be manually adjusted using the keyboard 30 to optimize car-
diac output. This is advantageously accomplished while the

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system is providing circulatory assist and cardiac output is being monitored and displayed. After such manual optimization, it is appropriate to re-enter the program starting at step 108.

If the automatic mode of operation is selected, then branching from step 120 is to step 122, where a subroutine to automatically optimize delay time t_d, described below with respect to FIG. 7, is initiated. After the delay is optimized a subroutine which optimizes the inspiration pulse width t_{ip}, as described below with respect to FIG. 8, is executed at step 124. The optimum values of the delay time t_d, the pulse width t_{ip} and cardiac output Q_n are stored at 126.

The parameter which has been observed to be most critical in determining the extent of the cardiopulmonary circulatory assist provided is the delay time t_d optimized at step 122. However, adjustment of the pulse width t_{ip} at step 124 may also have an influence on cardiac output. Thus, it is advantageous to adjust the delay and width iteratively. This is performed by providing an adjustment loop. At step 128 a counter, initialized at a value of 0, is incremented by one count. At step 130 a determination is made as to whether the count is less than a predetermined integer Y which may have a value of, for example, two or three. Assuming that the present count at step 128 is less than the value of Y, the program loops back to step 122. The delay t_d is then optimized at 122, the width t_{id} is again optimized at 124 and the new optimum values including cardiac output, are stored at 126. The count is incremented at step 128. When the count is equal to Y, the program branches from step 130 to step 132 where the counter is reset. At step 134 cardiac output is continuously monitored to determine whether it has decreased below 0.9 times the optimum value stored at step 126. If this has occurred, the system begins at step 136 to log data into RAM 16. The data can

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also be stored on a disk in a disk drive. The data may
include pulse rate, the output of the continuous cardiac
output monitor 66 or any other relevant physiologic data,
which is then available for later interpretation and review
by the physician.

At step 138 a determination is made as to whether
cardiac output has decreased below 0.7 times the maximum
value Q_{nMAX} (the maximum of the values stored at step 126)
which occurred since circulatory assist was commenced,
rather than the most recently stored value. As long as the
answer is NO, the program loops back to step 122 and a
reoptimization occurs. However, if the answer is YES an
alarm is activated at step 140 prior to looping to step 122,
since a decrease of more than 30% in cardiac output is con-
sidered sufficient cause to warrant investigation, rather
than simply logging data.

Referring to FIG. 7 the delay optimization sub-
routine of step 122 is described. After initializing appro-
priate registers associated with the subroutine when start-
ing at step 142, the current value of cardiac output Q_n is
stored at step 144. At step 146 an initial value of the
delay time t_{id} is incremented by ten milliseconds to provide
a new value. The initial value may be manually entered at
step 108 or may be a default value. In either case, this
initial value should be chosen so as to be less than that
anticipated as optimum. If this is the case, when the time
delay is incremented at step 146 there will be an increase
in cardiac output which is then provided as input Q_{n+1} at
step 148. At step 150 a comparison between the cardiac
output Q_n at step 144 and the cardiac output Q_{n+1} at step
148 is made after a delay of at least two minutes to allow
for establishing a new state of equilibrium. As long as the
cardiac output has not decreased, the program loops to step
152 where the new value of Q_n is set equal to that deter-
mined at step 148. Again the delay time is incremented at

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step 146 and the latest value of the cardiac output is
accessed at step 148. The determination at step 150 is
again made. The loop including steps 146, 148, 150 and 152
continues until an increment causes the cardiac output to
drop slightly. At this time the program exits from the
subroutine. Although not shown, it is possible, if desired,
to decrement the time delay by ten milliseconds just prior
to exiting.

Referring to FIG. 8 a similar routine is followed
with respect to the pulse width t_{ip}. After starting at step
156, the current cardiac output Q_n is provided as an input
at step 158. The initial inspiration plateau time t_{iip}
(initial pulse width or inspiration time) is incremented by
a fixed value I such as ten milliseconds. At step 162 a new
value for the cardiac output Q_{n+1} is accessed. At step 164
a determination similar to that at step 150 is made. If the
cardiac output has not decreased then the program branches
to step 166 where the current value of cardiac output Q_{n+1}
is read in as the new value of Q_n. The pulse width is again
incremented at step 160 and the program proceeds to step 162
and to step 164. The loop including steps 160, 162, 164 and
166 continues until such time as the cardiac output is
determined, at step 164, to have decreased relative to the
previous value. The subroutine is then exited at 166. As
noted above with respect to FIG. 7, before exiting from the
subroutine the inspiration pulse width may be decremented by
an amount equal to the previous increment so that the
cardiac output is maximized. However, in view of the
relatively small changes being made, this step is not
regarded as being critical, particularly with respect to the
pulse width.

Various modifications of the invention will become
apparent to those skilled in the art. For example, although
not described herein it will be apparent that automatic
selection of the value of N may be implemented. Thus, if

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the patient's heartbeat increases above some predetermined
rate, it may be desirable to change from a 1:1 mode to a 1:2
mode. A lookup table of heart rate versus N value may be
provided. However, it will be recognized that some hystere-
sis must be built in so that the mode change is not rapid
and repetitive so as to produce discomfort in the patient.
Further, as noted above, if the mode is changed, a change in
the pulse width t_{ip} must occur as well in order to provide
adequate ventilation if the apparatus of the invention is
interfaced to the patient using the arrangement of FIG. 2A
or FIG 3.

The present invention is particularly well suited
and most effective in providing cardiopulmonary circulatory
assistance in patients having marginal circulation. In
fact, the more difficulty the patient is experiencing, the
greater will be the assist provided by the invention. In
this regard the invention may be of critical importance in
prolonging the lives of patients undergoing treatment for
severe circulatory disorders.

Although the invention has been described with
reference to a particular embodiment, it is to be understood
that this embodiment is merely illustrative of the applica--
tion of the principles of the invention. Numerous modifica-
tions may be made therein and other arrangements may be
devised without departing from the spirited scope of the
invention. For example, while the invention is advanta-
geously implemented using a microcomputer, a manually con-
trolled system may be implemented using discrete circuitry,
including delay generators and a divide by N circuit. In
addition, a hybrid approach, wherein such circuits are con-
trolled by a microcomputer may also be implemented. Other
equivalent approaches will occur to those skilled in the art.

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CLAIMS

1. A method for providing circulatory assistance
to a patient comprising the steps of:
        detecting the onset of ventricular ejection
in the cardiac cycle of the patient, and
        selectively increasing intrathoracic pressure
of the patient in relative phase with respect to said ven-
tricular ejection at a time which reduces ventricular after-
load and maintains ventricular preload.

2. The method of claim 1, further comprising the
steps of:
        monitoring cardiocirculatory output, and
        adjusting said relative phase to maximize
said output.

3. The method of claim 1, wherein said step of
selectively increasing said intrathoracic pressure is
performed once every N cardiac cycles, where N is a positive
integer.

4. The method of claim 3, wherein said positive
integer is one or two.

5. The method of claim 1, wherein said intratho-
racic pressure is increased by at least one of:
        a) injecting air into the patient's lungs;
and
        b) exerting external pressure on the thorax
of the patient.

6. The method of claim 1, wherein said exerting
of external pressure is achieved by increasing the pressure
of fluid in a bladder surrounding the thorax of the patient.

7. The method of claim 1, further comprising the
steps of:
        monitoring cardiocirculatory output; and
        adjusting duration of a time interval for
which said increasing of intrathoracic pressure occurs to

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maximize said output.

     8.   The method of claim 7, further comprising the step of adjusting said relative phase to maximize said output.

     9.   The method of claim 8, further comprising repeatedly performing the steps of adjusting said duration and adjusting said relative phase to maximize said output.

     10.   The method of claim 7, wherein said step of monitoring said cardiocirculatory output comprises:

     determining oxygen uptake by the patient;

     determining a difference in arterial oxygen concentration and venous oxygen concentration;

     dividing the oxygen uptake by said difference to determine said cardiocirculatory output.

     11.   The method of claim 1, wherein said introthoracic pressure is selectively increased during isovolumetric contraction of the heart.

     12.   An apparatus for improving the cardiac output of a patient, said apparatus comprising:

     detection means for detecting the onset of ventricular ejection in the cardiac cycle of the patient;

     pressure increasing means for selectively increasing intrathoracic pressure of the patient;

     control means responsive to said detection means for activating said pressure increasing means in relative phase with respect to said ventricular ejection;

     cardiocirculatory monitoring means for ascertaining the extent to which said apparatus improves cardiac output and for providing an output signal indicative thereof; and

     delay means responsive to said output signal, said delay means being for delaying activation of said control means in response to said detecting means so that said cardiac output is maximized.

     13.   The apparatus of claim 12, wherein said car-

diocirculatory monitoring means includes means for determin-
ing a rate of flow of blood from the heart.

        14. The apparatus of claim 12, wherein said
cardiocirculatory monitoring means includes an oxygen
consumption monitoring means for determining the rate of
oxygen consumption of the patient.

        15. The apparatus of claim 14, wherein said
oxygen consumption monitoring means comprises:

            means for supplying an oxygen stream to the
patient,

            first concentration measuring means for
measuring concentration of oxygen in said oxygen stream;

            second concentration measuring means for
monitoring concentration of oxygen in said stream at a point
including gases exhausted by the patient; and

            flow measuring means for measuring flow rate
of gases in said stream.

        16. The apparatus of claim 14, further comprising
blood oxygen measuring means for measuring the difference in
arterial oxygen concentration and venous oxygen concentra-
tion of the patient and for providing a difference output
indicative of said difference in arterial oxygen concentra-
tion and venous oxygen concentration.

        17. The apparatus of claim 16, further comprising
calculating means for determining cardiac output; said
calculating means including:

            difference means for determining the differ-
ence in oxygen concentration measured by said first concen-
tration measuring means and said second concentration mea-
suring means;

            multiplying means for multiplying said
difference by said flow rate measured by said flow measuring
means to determine a value of said instantaneous oxygen
consumption; and

            dividing means for dividing said value of

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instantaneous oxygen consumption by said difference output
of said blood oxygen measuring means to determine a cardiac
output value.

18. The apparatus of claim 17, further comprising
integrating means for integrating values of said cardiac
output to determine an averaged cardiac output.

19. The apparatus of claim 15, wherein said
integrating means comprises:

a first in/first out memory for storing said
values of cardiac output; and

means for determining the sum of values in
said memory.

20. The apparatus of claim 12, further comprising
divider means responsive to said detection means, said
divider means providing an output every N cardiac cycles,
where N is a positive integer, said control means being
responsive to said output.

21. The apparatus of claim 20, further comprising
selection means for selecting said positive integer.

22. The apparatus of claim 20, wherein said
positive integer may be selected by said selection means to
be one or two.

23. The apparatus of claim 12, wherein said
pressure increasing means comprises:

a source of pressurized fluid;

pressure pulse means for providing pulses of
said pressurized fluid; and

a patient interface means for utilizing said
pulses to increase said intrathoracic pressure.

24. The apparatus of claim 23 wherein said
patient interface means includes a bladder in contact with
the thorax of the patient.

25. The apparatus of claim 24, where said bladder
is shaped and sized to surround the thorax of the patient.

26. The apparatus of claim 23, wherein said

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pressurized fluid is one of air and an air/oxygen mixture
and said patient interface means includes at least one of a
tube for supplying said pressure pulses to the lungs for
inspiration and an inflatable bladder in contact with the
thorax of the patient.

     27. The apparatus of claim 26, wherein said blad-
der is sized and shaped to surround at least a substantial
portion of the thorax of the patient.

     28. An apparatus for improving cardiac output of a
patient, said apparatus comprising:

     detection means for detecting onset of
ventricular ejection in the cardiac cycle of a patient;

     pressure increasing means for selectively
increasing intrathoracic pressure of the patient; and

     control means responsive to said detection
means for activating said pressure increasing means in
relative phase with respect to said ventricular ejection at
a time which reduces ventricular afterload and maintains
ventricular preload.

     29. An apparatus for determining cardiac output of
a patient comprising:

     means for measuring rate of oxygen consumption
of the patient;

     means for measuring a difference in arterial
oxygen concentration and venous oxygen concentration in the
patient and for providing a difference value representative
of said difference; and

     divider means for dividing said rate of oxygen
consumption by said difference value to determine said car-
diac output.

     30. The apparatus of claim 29, wherein said
means for measuring oxygen consumption of the patient
comprises:

     means for supplying an oxygen stream to the
patient;

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first concentration measuring means for mea-
suring concentration of oxygen in said oxygen stream;

second concentration measuring means for
monitoring concentration of oxygen in said stream at a point
including gases exhausted by the patient; and

flow measuring means for measuring flow rate of
gases in said stream.

31. The apparatus of claim 30, further comprising
calculating means for determining instantaneous oxygen
consumption, said calculating means including:

difference means for determining a difference
in oxygen concentration measured by said first concentration
measuring means and said second concentration measuring
means; and

multiplying means for multiplying said differ-
ence by said flow rate measured by said flow measuring means
to determine a value of said instantaneous oxygen consump-
tion.

32. The apparatus of claim 31, further comprising
integrating means for integrating values of said cardiac
output to determine an averaged cardiac output.

33. The apparatus of claim 32, wherein said
integrating means comprises:

a first in/first out memory for storing said
values of cardiac output; and

means for determining a sum of values in said
memory.

34. An apparatus for determining rate of oxygen
consumption of a patient, comprising;

means for supplying an oxygen stream to the
patient;

first concentration measuring means for
measuring concentration of oxygen in said oxygen stream;

second concentration measuring means for
monitoring concentration of oxygen in said stream at a point

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including gases exhausted by the patient; and

        flow measuring means for measuring the flow rate of gases in said stream.

        35. The apparatus of claim 34, further comprising calculating means for determining instantaneous oxygen consumption, said calculating means including:

        difference means for determining the difference in oxygen concentration measured by said first concentration measuring means and said second concentration measuring means; and

        multiplying means for multiplying said difference by said flow rate measured by said flow measuring means to determine a value of said instantaneous oxygen consumption.
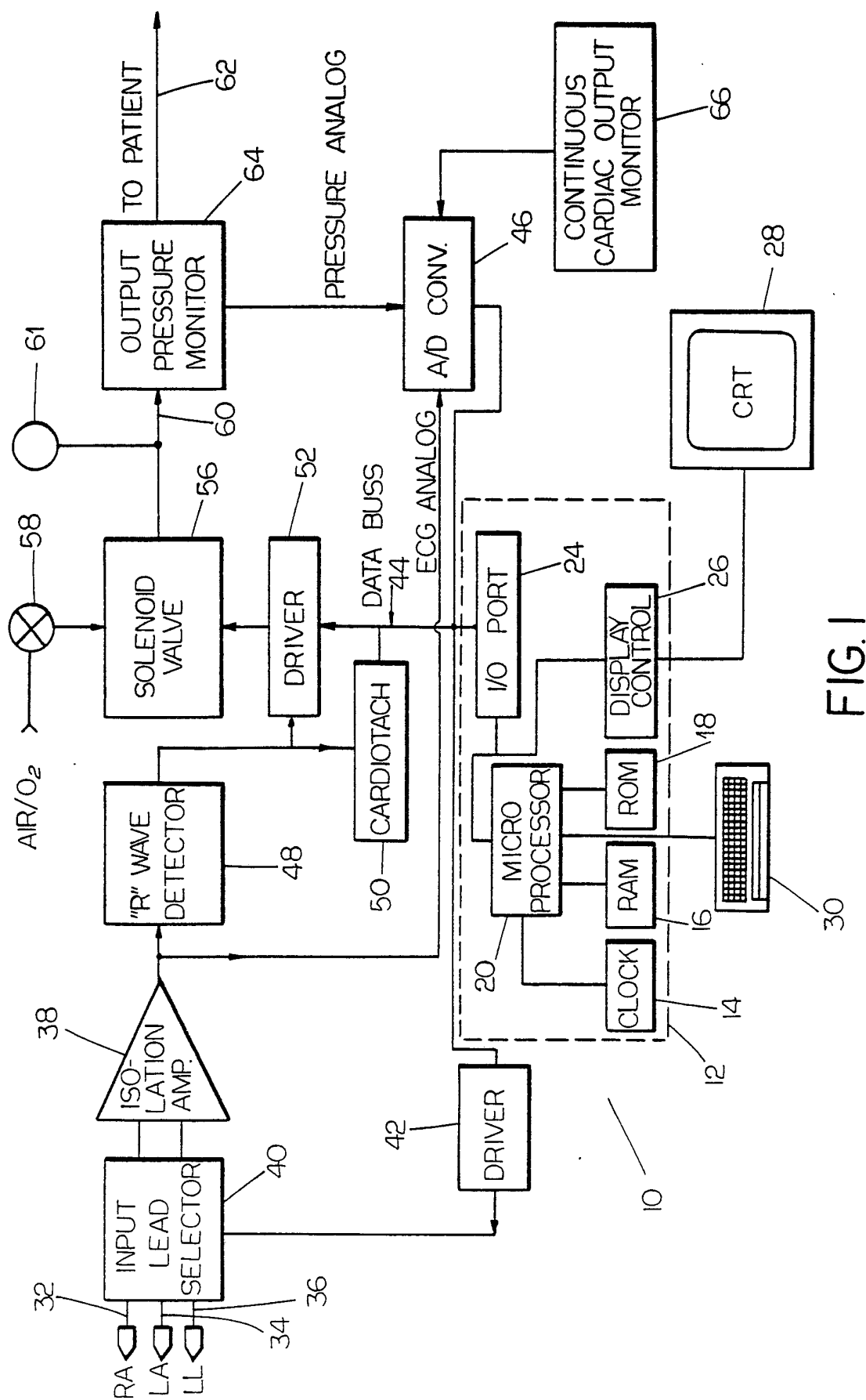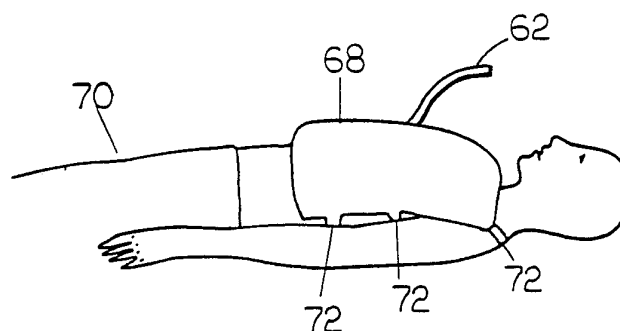
FIG. 1

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FIG. 2
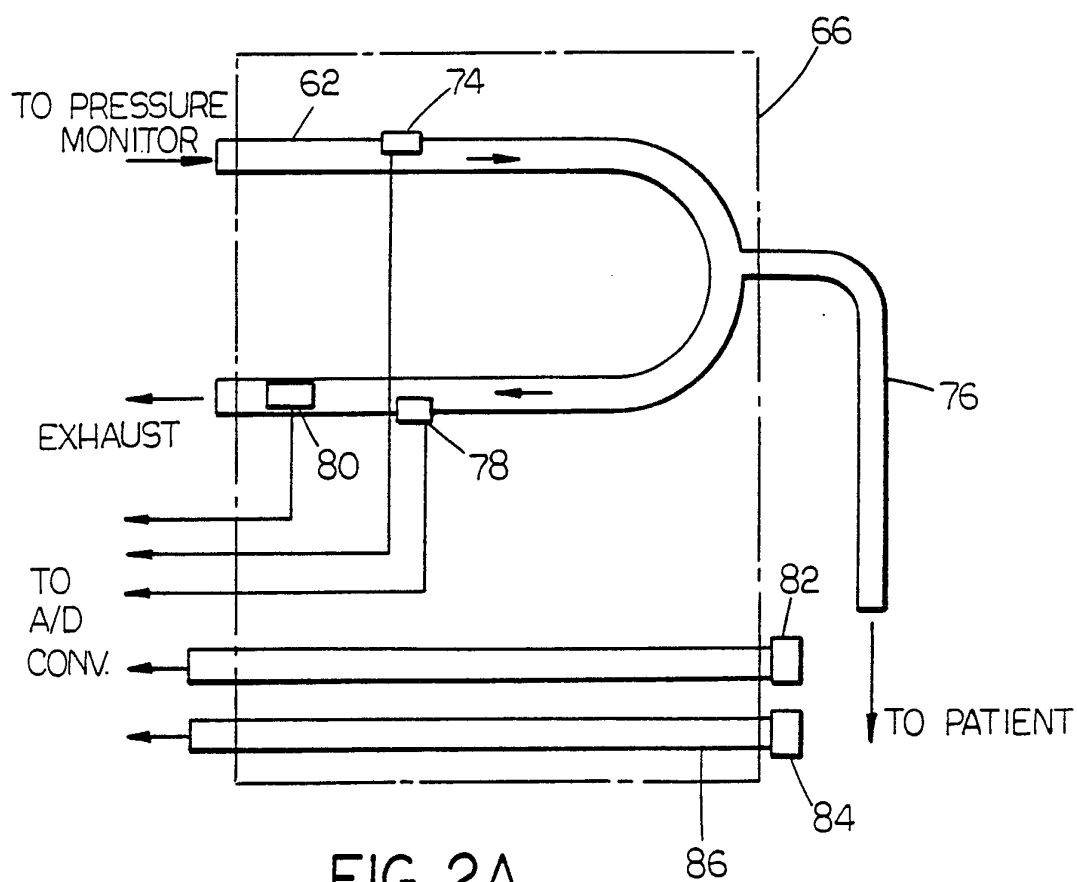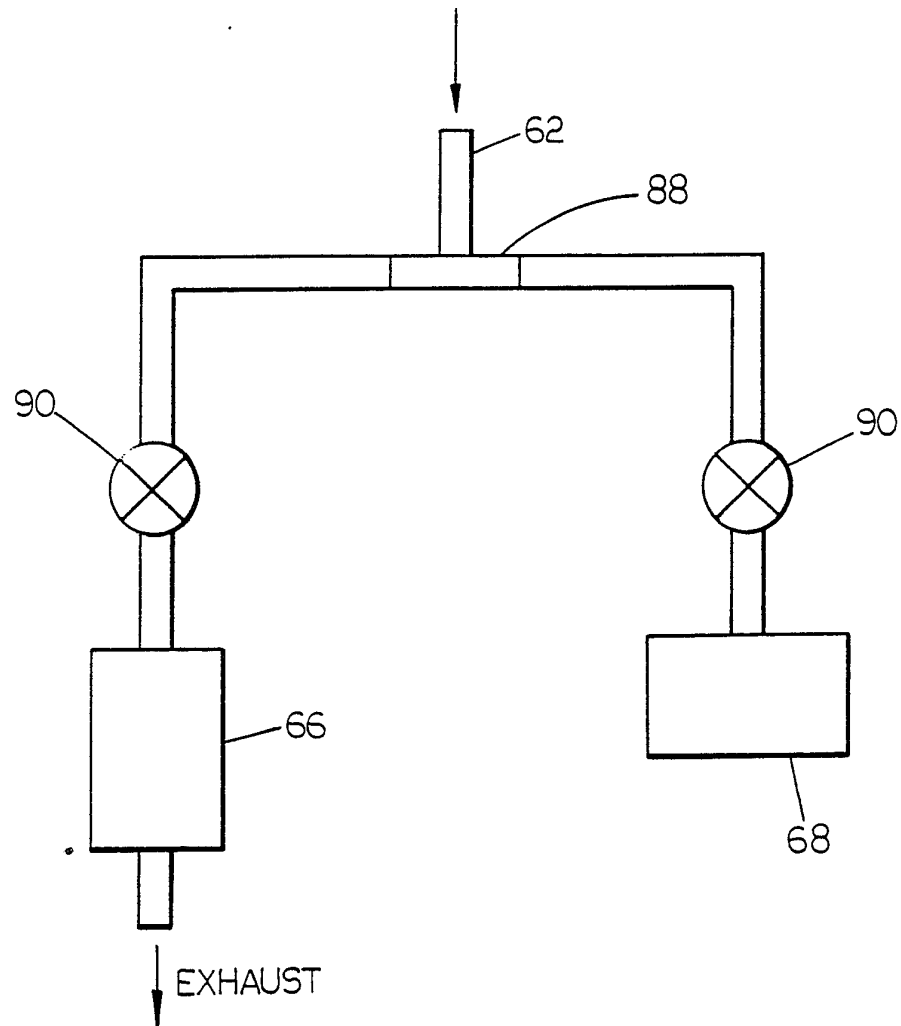

FIG. 2A

FIG.3

SPONTANEOUS
CIRCULATION

ASSISTED
CIRCULATION

FIG.4A

Ao

FIG.4B

ECG

t_d t_{ip}

FIG.4C

SOLENOID
SQUARE WAVE

FIG.4D

Q_V

FIG.4E

Q_F

FIG. 5A   Aₒ

FIG. 5B   ECG

FIG. 5C   SOLENOID
                                                                     SQUARE WAVE

FIG. 5D   Qᵥ

FIG. 5E   Q_F

FIG. 6A

Flowchart:

START — 100

INPUT PATIENT DATA, DATE, ETC. — 102

SWITCH TO PHYSIOLOGIC DISPLAY — 104

BEGIN C. C. O. LOOP — 106

ENTER CONTROL DATA/MODE — 108

START ASSIST — 110

15 SECOND DELAY — 112

$Q_F \leq Q$ — 114

YES → STOP ASSIST — 116 → ALARM — 118

TO 120

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FROM 114

AUTO MODE SELECT-ED — 120

NO

$Q_F < 0.7 Q_F MAX$ — 121   YES   TO 118

NO

YES

OPTIMIZE DELAY — 122

OPTIMIZE WIDTH — 124

STORE OPTIMUM WIDTH, DELAY Q_F — 126

COUNT TIMES IN LOOP. — 128

RESET COUNTER — 132

NO

$Q_F < 0.9 Q_F OPT.$

YES   136

BEGIN CONTINUOUS DATA LOG

138

$Q_F < 0.7 Q_F MAX$ NO

YES   140

ALARM

YES   COUNT < Y   NO

130

FIG. 6B

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START — 142

INPUT Q_n — 144

FIG. 7

$t_d = t_{1d} + 0.01\,sec.$ — 146

$Q_{n+1} = Q_n$ — 152

148 — INPUT Q_{n+1}

154

RETURN   YES   150
$Q_{n+1} : Q_n$
$Q_{n+1} = $MAX   NO

START — 156

INPUT Q_n — 158

166

FIG. 8

$t_{1p} = t_{11p} + I$ — 160

$Q_{n+1} = Q_n$ — 166

INPUT Q_{n+1} — 162

166

RETURN   YES   164
$Q_{n+1} : Q_n$
$Q_{n+1} = $MAX   NO

SUBSTITUTE SHEET

INTERNATIONAL SEARCH REPORT

I. CLASSIFICATION OF SUBJECT MATTER (if several classification symbols apply, indicate all) 6

According to International Patent Classification (IPC) or to both National Classification and IPC

IPC(4) A61H 31/00
U.S. Cl. 600/17

II. FIELDS SEARCHED

Minimum Documentation Searched 7

Classification System	Classification Symbols
U.S.	600/16,17 128/671,695,696,700,725,30.2,28,24R,204.23, 204.24,64

Documentation Searched other than Minimum Documentation
to the Extent that such Documents are Included in the Fields Searched 8

III. DOCUMENTS CONSIDERED TO BE RELEVANT 9

Category *	Citation of Document, 11 with indication, where appropriate, of the relevant passages 12	Relevant to Claim No. 13
Y	US,A, 3,835,845 (MAHER) 17 September 1974, See entire document.	1,2,7-9,12,23
Y	US,A, 3,266,487 (WATKINS et al.) 16 August 1966, col. 9, lines 50-64, col. 11, lines 32-49.	3,4,22
Y	US,A, 3,426,743 (CHESTNUT et al.) 11 February 1969, col. 21, lines 30-65.	3,4,22
Y	US,A, 3,750,644 (RAGSDALE) 07 August 1973, See entire document.	1,2,7-9,12
Y	US,A, 3,985,123 (HERZLINGER et al) 12 October 1976, See entire document.	2,7-10,12-17, 29-31,34,35
Y	US,A, 3,966,358 (HEIMES et al.) 29 June 1976, See entire document.	2,7-10,12-17, 29-31,34,35
Y	US,A, 3,923,055, (HAMMACHER) 02 December 1975, See entire document.	1-5,7,11,12,15, 23,30,34
Y	US,A, 4,077,402, (BENJAMIN, Jr. et al.) 07 March 1978, See entire document.	2,7-9,12,13
Y	US,A, 4,204,524, (MARTIN et al) 27 May 1980, See entire document.	1-28

* Special categories of cited documents: 10

"A" document defining the general state of the art which is not considered to be of particular relevance

"E" earlier document but published on or after the international filing date

"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)

"O" document referring to an oral disclosure, use, exhibition or other means

"P" document published prior to the international filing date but later than the priority date claimed

"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step

"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art.

"&" document member of the same patent family

IV. CERTIFICATION

Date of the Actual Completion of the International Search	Date of Mailing of this International Search Report
30 June 1989	2 5 JUL 1989
International Searching Authority	Signature of Authorized Officer
ISA/US	Kennedy J. Schaetzle

FURTHER INFORMATION CONTINUED FROM THE SECOND SHEET	
Y US,A, 3,078,842,((GRAY), 29 June 1959, See entire document.	1,5,7,23-28
Y US,A, 4,058,855, (RUNGE), 22 November 1977, See entire document.	1,2
Y US,A, 4,676,232, (OLSSON et al.) 30 June 1987, See entire document.	1-28

V. ☐ OBSERVATIONS WHERE CERTAIN CLAIMS WERE FOUND UNSEARCHABLE [1]

This international search report has not been established in respect of certain claims under Article 17(2) (a) for the following reasons:

1. ☐ Claim numbers _____ , because they relate to subject matter [12] not required to be searched by this Authority, namely:

2. ☐ Claim numbers _____ , because they relate to parts of the international application that do not comply with the prescribed requirements to such an extent that no meaningful international search can be carried out [13], specifically:

3. ☐ Claim numbers _____ , because they are dependent claims not drafted in accordance with the second and third sentences of PCT Rule 6.4(a).

VI. ☐ OBSERVATIONS WHERE UNITY OF INVENTION IS LACKING [2]

This International Searching Authority found multiple inventions in this international application as follows:

1. ☐ As all required additional search fees were timely paid by the applicant, this international search report covers all searchable claims of the international application.

2. ☐ As only some of the required additional search fees were timely paid by the applicant, this international search report covers only those claims of the international application for which fees were paid, specifically claims:

3. ☐ No required additional search fees were timely paid by the applicant. Consequently, this international search report is restricted to the invention first mentioned in the claims; it is covered by claim numbers:

4. ☐ As all searchable claims could be searched without effort justifying an additional fee, the International Searching Authority did not invite payment of any additional fee.

Remark on Protest

☐ The additional search fees were accompanied by applicant's protest.

☐ No protest accompanied the payment of additional search fees.