THE INTERRELATIONSHIP OF FACTORS CONTROLLING CARDIAC OUTPUT

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ABSTRACT

Our understanding of the mechanics of circulation may be broadened
by testing old and new concepts on a new hydraulic model made possible
by a unique pump which resulted from the evolution of open heart surgi-
cal equipment. Findings in human physiology which have corollaries in
the model can be analyzed more easily in the model, with the resulting
conclusions transferable with reasonable validity. Factors thought to
control cardiac output were tested. The results indicate that there
are two separate and distinct sets of control factors, with only one
set being operative at a time. It is the set that is potentially lim-
iting cardiac output the most that is the determinant at any time.
Past paradoxes in our perception of human circulatory physiology are
accommodated by the resulting concept.

Key Words: Cardiovascular Dynamics, Cardiovascular Function, Circula-
tion Rate Control, Cardiac Output, Mean Vascular Pressure.

INTRODUCTION

A hydraulic model of the cardiovascular system, upon which new and
old physiologic concepts about circulation may be tested, has been a
by-product of the development of cardiac bypass equipment used in
surgery.

The duplication of many physiologic cause and effect phenomena is
made possible in the model by an unique pump which has closely mimicked
heart function in a large number of animal experiments [1] and human
clinical applications [2].

Most variables and reflexes thought to effect cardiac output can be
tested in the model. Study of the model has the advantage over study-
ing "in vivo" preparations as one or more variables at a time can be

changed in the model without the uncertainty that some hidden, unde-
tected change might have occurred that would invalidate conclusions.

Study of the hydraulic model also has the advantage over study of a
computer or mathematical model, as fluid mechanics eliminates the possi-
bility of overlooked or exaggerated evaluation of factors in a very
complex hydraulic system.

Findings using the model have resulted in a concept of cardiac out-
put regulation that accommodates paradoxes in explaining:

1) Why pulse rate alteration does not always effect cardiac output:
   a) When altering pacemaker rates in patients with non-failing
      hearts (3,4).
   b) When decreasing pulse rates with propranolol in normal sub-
      jects (5).
   c) When increasing pulse rates during hypovolemic, neurogenic
      and endotoxic shock (6,7,8,9).

2) Why strength of myocardial contraction alteration does not
   always alter cardiac output:
   a) When digitalis is given to normal subjects (10).
   b) When propranolol decreases myocardial contractility in non-
      failure states (5,11).
   c) When intra-aortic balloon augmenting myocardial contraction
      causes no increase in cardiac output in non-failure states
      (12).

3) Why total vascular resistance increase causing essential hyper-
   tension is not accompanied by a decrease in cardiac output (13,
   14).

4) Why "afterload" increase from aortic stenosis (15) or coarctation
   of the aorta (16) is unaccompanied by a decrease in cardiac out-
   put.

5) Why "pre-load" increase during heart failure fails to increase
   cardiac output.

THE MODEL

Figures 1 and 2 show the major components of the simulated cardio-
vascular system.

Pumps: The unique simulator heart pumps share three characteristics of
the heart that allow them to closely approximate cardiac function (1,2):
they are non-sucking at their inlets, allow uninterrupted venous flow to
their inlets, while providing pulsatile outflow to the arteries.

Each pump consists of an atrial chamber and a ventricular chamber
with inlet and outlet ball valves (Fig. 3). These pumping chambers have
rigid Lucite saucer-shaped outer shells (Fig. 4A) with mirror image rub-
ber diaphragm inner shells (Fig. 4B). These pumping chambers are opposed
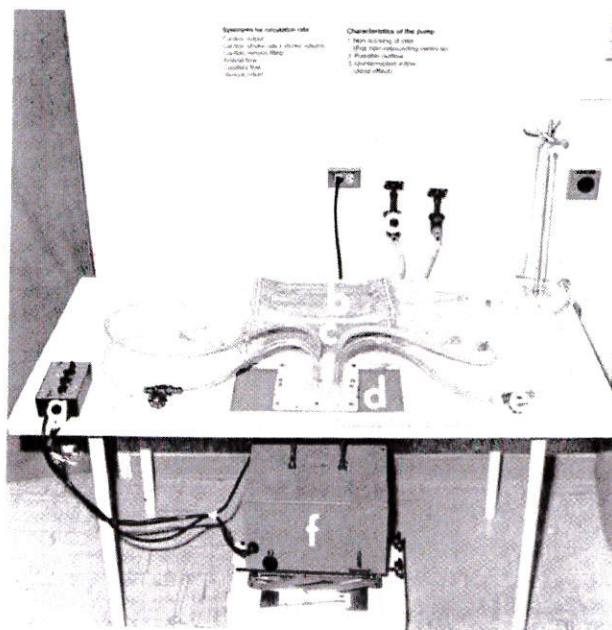
Figure 1. Circulation Model:
a - Compressed air and suction sources
b - Simulated systemic vascular bed
c - Simulated pulmonary vascular bed
d - Right and left heart pumps
e - Metering valve
f - Solonoid valves and suction and pressure tanks
g - Electronic timing device

by similar shaped impeller chambers which also consist of rigid Lucite outer shells (Fig. 4E) and inner rubber diaphragms (Fig. 4D). A 1/4-inch thick Lucite gasket (Fig. 4C) placed between the opposing impeller and pumping chambers has side holes that make free communication between the room air and the space between the opposing diaphragms.

The pumps are driven by alternately delivering compressed air and suction to the impeller chambers. An electronic regulator triggers solonoid valves to control the pulse rate and ratio of systole to diastole. Reducer valves regulate peak pressure and onset to peak time.

1) Non-sucking characteristic: The Lucite gasket with side holes lets room air between the pumping and impeller diaphragms during forceful evacuation of the impeller chamber during ventricular diastole. The negative pressure in the impeller chamber is thus applied to the room air instead of to the pumping diaphragm.

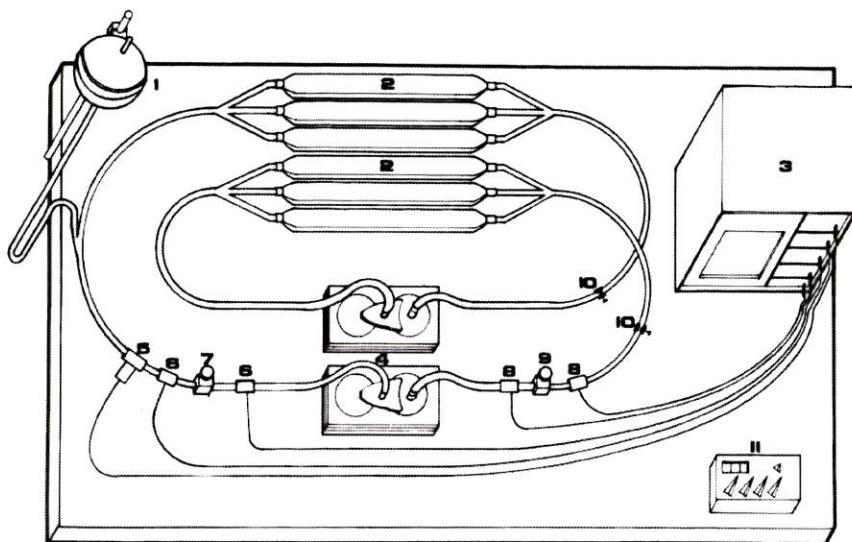2) Uninterrupted inflow to the pumps is provided by the atria.

Figure 2. Circulation Model:
  1 - Transfusion reservoir
  2 - Simulated systemic and pulmonary vascular beds
  3 - Electromagnetic flow meter and pressure recorder
  4 - Simulated right and left heart non-sucking pumps
  5 - Flow probe
  6 - Inflow pressure transducers
  7 - Inflow resistance metering valve
  8 - Outlet pressure transducers
  9 - Outlet resistance metering valve
  10 - Additional movable resistance clamps
  11 - Electronic timing controls

Atria emptied during ventricular diastole accommodate venous flow during
ventricular systole, thus preventing interruption of flow during ven-
tricular systole. Then during ventricular diastole, atria empty the
accumulated blood into the ventricles without venous interruption. A
rigid "Y" segment of the atrium between the inlet veins and the inlet
valves prevents interruption of venous flow by the blocking of a com-
pletely emptied atrium that would otherwise occur at the end of atrial
systole.

   3) Pulsatile outflow is provided by alternate positive and negative
air pressure being delivered to the impeller chambers. The pulse rate
is variable from 1 to 160 compressions/min. with a pressure from 1 to
30 p.s.i.

The Simulated Vascular Network consists of a systemic and pulmonary net-
work in series with the two pumps interposed. Major vessels are
silicone tubing, while the capillary beds are made of more compliant
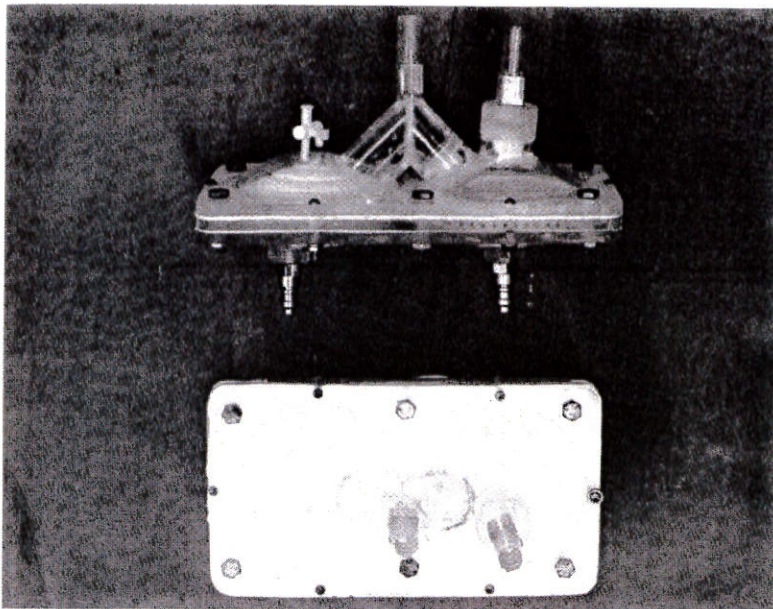Penrose drain material.

Figure 3. Assembled model heart pump, side and top views.

Metering valves for altering resistance are incorporated in the venous and arterial tubing (Fig. 1,e). Pressure-sensitive transducers (Fig. 2, #5, #6 & #8), connected to a direct writing recorder and electromagnetic flowmeter and probes, are used to monitor pressures and flow. A "transfusion" reservoir permits alteration in the fluid volume within the system. Weights, placed on a plate resting on the capillary and venular portions of the model, are used to vary the compliance of the system.

PROCEDURES

The effect on pump output from modifying one circulation variable at a time, while keeping others constant, was measured using the model. The variables tested were: 1) mean vascular pressure, 2) pump rate, 3) strength of pump contraction, and 4) resistance in the vascular network.

Experiment #1. Relationship of Mean Vascular Pressure to Pump Output.

("Mean vascular pressure" is the resting pressure in the system that results from the fluid volume and the compliance of the system. Mean vascular pressure was measured with the pumps turned off with the pressure equalized in all parts of the model).

Mean vascular pressure was changed by adding fluid, leaving compliance, the other factor, unchanged. Starting with a mean vascular pres-
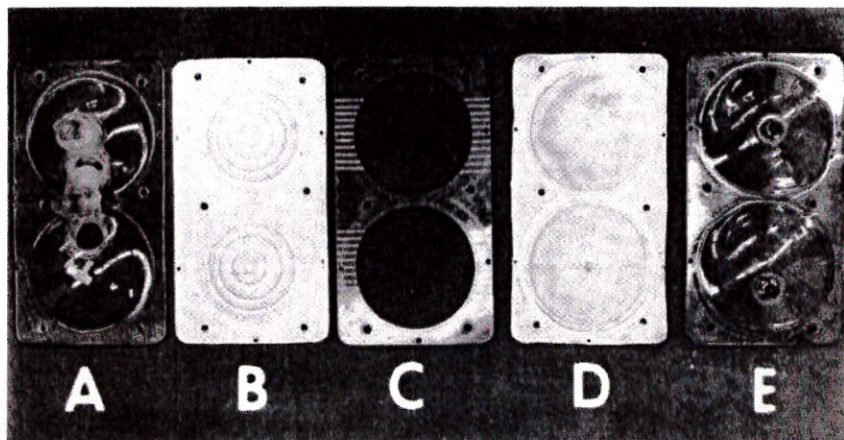
Figure 4.   Disassembled model heart pump.
A - Atrial and ventricular pumping chamber shell and valves.
B - Rubber pumping compartment diaphragm.
C - Spacer with side holes which allow free movement of ambient air
    to the space between the impeller and pumping diaphragms.
D - Silicone rubber impeller diaphragm.
E - Impeller chambers with atrial and ventricular connectors for
    suction and compressed air.

sure at zero, with 2,000 cc. volume in the model, appropriate amounts
of fluid were added to change the mean vascular pressure in steps of
2 cm. water pressure. The amounts added were predetermined from a vol-
ume pressure curve made while adding fluid to the system with the pumps
off (Fig. 5). Factors kept constant besides compliance were pump rate
of 80 beats/min., impeller pressure of 16 p.s.i. and the resistance.

Findings:

1) Figure 6 shows a direct correlation between pump output and mean
vascular pressure over the range of 0 to 20 cm. of water pressure.
2) At no time during the range of pressure studied did the ventricles
fill maximally at diastole.
3) At no time was there any interruption in venous flow by the in-
termittent closing of the inlet valves.

Experiment #2.   Pump Rate Correlation with Pump Output.

The pump rate in the model was varied from 0 to 170 beats/min. with
other variables remaining constant (Fig. 7). The mean vascular pressure
was at 12 cm. water pressure, with the impeller pressure at 14 p.s.i.
and the resistance kept constant.

Findings:

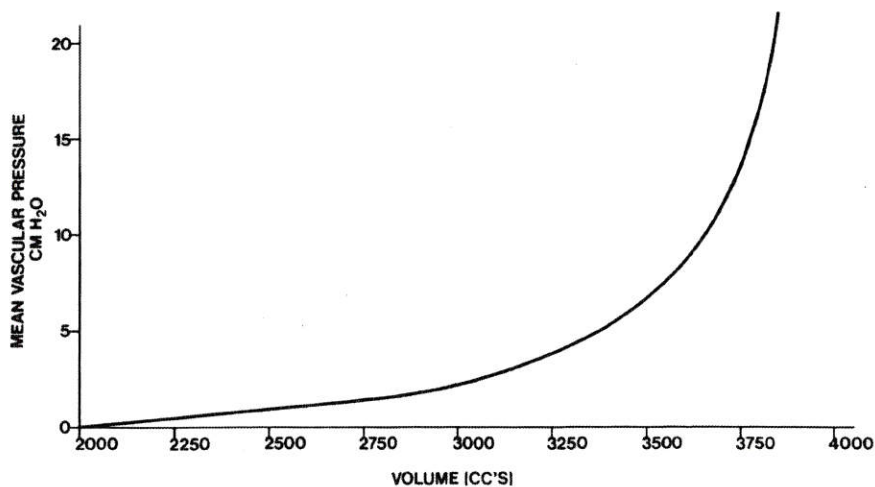1) At pump rates between 0 and 20/min.:

Figure 5.   Mean vascular pressure - Volume relationship in the circulation model.
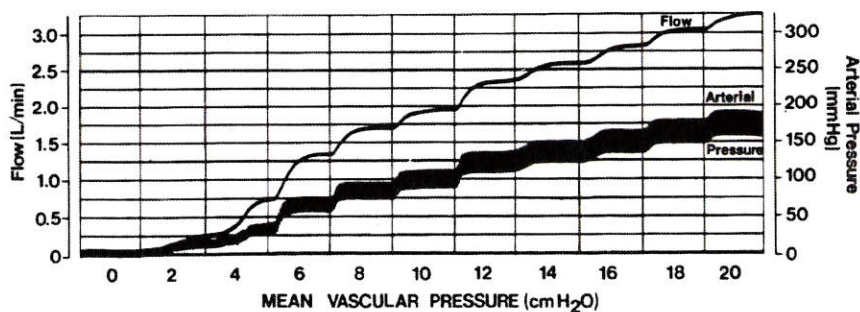

Figure 6.   Pump output correlation with mean vascular pressure. Such variations in mean vascular pressure result from either volume or compliance changes within the system.
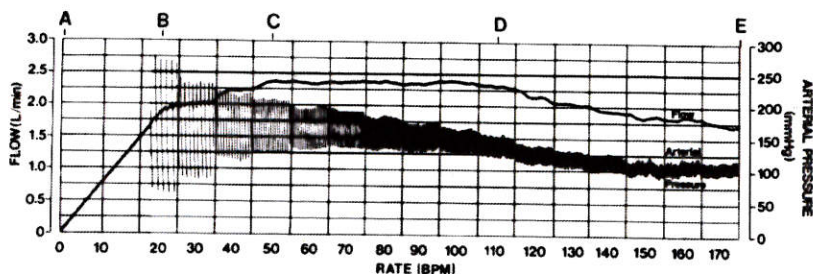
Figure 7. Relationship of Pump Rate to Flow. Composite tracing of flowmeter measurement above 20 beats/min. and mechanical measurement below 20 beats/min. where the flow-meter was non-operative.

a) The pump output correlated in a linear way with the pump rate (Fig. 7, A to B).
b) The ventricles filled to capacity at each ventricular diastole.
c) The atria filled to capacity during each ventricular systole.
d) Venous flow was interrupted during each ventricular systole.

2) At pump rates between 20 and 40/min.:
a) The pump rate correlated in a non-linear way with the pump output (Fig. 7, B to C).
b) The ventricles were filled slightly less than capacity at ventricular diastole.
c) The atria were filled to capacity at the end of ventricular systole.
d) Venous flow was slowed but not completely interrupted before the end of ventricular systole.

3) Increasing pump rate from 40 to 110:
a) Caused no increase in pump output (Fig. 7, C to D).
b) Was associated with progressive decrease in ventricular end-diastolic volume.
c) Caused progressive decrease in atrial end-diastolic volume.
d) Was unaccompanied by any interruption in venous flow.

4) Increasing the pump rate from 110 to 170 (Fig. 7, D to E) was accompanied by a progressive increase in residual air pressure in the ventricular impeller after ventricular systole which:
a) Progressively decreased pump output.
b) Impeded ventricular filling at ventricular diastole.
c) Caused venous flow interruption at ventricular systole.

Experiment #3. Ventricular Impeller Force Correlation with Pump Output.

The ventricular impeller pressure was varied from 0 to 20 p.s.i., in increments of 2 p.s.i., while keeping other variables constant. The

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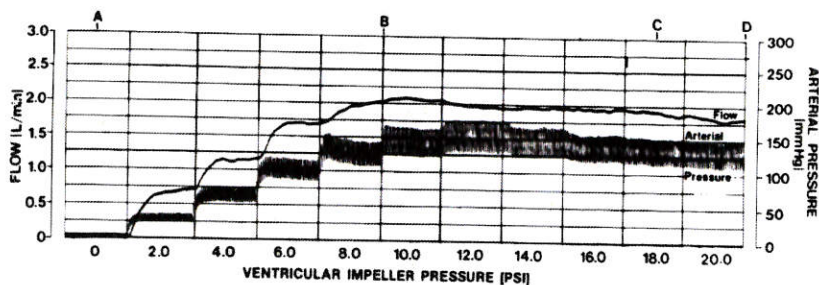pump rate was 80, the mean vascular pressure was 12 cm. water pressure and all resistance remained unchanged.

Figure 8.   Relationship between ventricular impeller
pressure and pump output.

Findings:

1) At impeller pressures between 0 and 10 p.s.i. there was:
   a) Linear correlation between impeller pressure and pump output
      (Fig. 8, A to B).
   b) The ventricles were never completely emptied at ventricular
      systole.
   c) The ventricles were completely filled at ventricular
      diastole.
   d) Venous flow to the pumps was interrupted at each ventric-
      ular systole.

2) At impeller pressures between 10 and 18 p.s.i. there was:
   a) No increase in the output as the pressure increased
      Fig. 8, B to C).
   b) Maximal ventricular emptying at systole.
   c) Sub-maximal filling of the ventricles at diastole.
   d) No venous flow interruption at any time.

3) As the impeller pressure was progressively increased above
   18 p.s.i. (Fig. 8, C to D) there was progressively incomplete
   evacuation of compressed air in the ventricle impeller during
   diastole which:
   a) Progressively decreased pump output.
   b) Impeded ventricular filling at diastole.
   c) Caused venous flow interruption at ventricular systole.

Experiment #4.   Relationship of Resistance to Pump Output

Resistance at a variety of sites was changed in the model's vascular
network while maintaining other factors constant with the mean vascular
pressure at 12 cm. water, pump rate at 80 and an impeller pressure at
16 p.s.i.   There was a marked difference in response to a given resis-
tance, depending upon where the resistance was placed in the model.

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Responses fell into two categories which became more obvious the closer the resistance was to either the inlet or outlet of a pump. Therefore, resistance was studied in two situations: near a pump outlet and near a pump inlet.

Experiment #4a.   Resistance Near a Pump Outlet.

A metering valve was placed 20 cm. from the left heart homologue pump outlet (Fig. 2, #9) across which a pressure gradient could be monitored by transducers (Fig. 2, #8), one on each side of the valve.
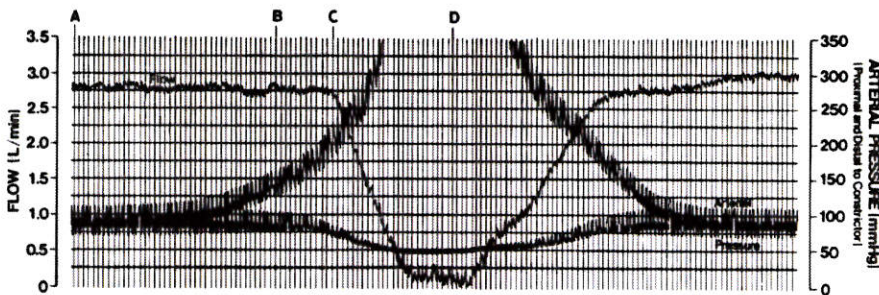

Figure 9.   Relationship of outflow resistance to pump output.
A - No resistance across outlet metering valve.
B - Beginning pressure gradient indicating addition of minimal resistance.
C - 100 mm Hg gradient manifesting moderate outflow resistance with normal flow.
D - Marked resistance with marked flow reduction.

Findings:

1) As the metering valve was progressively closed (Fig. 9) the superimposed arterial pressure tracings which demonstrated no gradient initially (Fig. 9, A to B), separated (Fig. 9, B to D) as resistance was added.

2) With increase in outflow resistance up to 100 mm Hg gradient there was no drop in pump output (Fig. 9, B to C).

3) Further increase in gradient from 100 to 250 mm Hg did cause a corresponding drop in flow (Fig. 9, C to D).

4) In the range where there was no drop in flow (Fig. 9, B to C) the left ventricle did not fill maximally, but did empty maximally, and there was no interruption of venous flow at any time during the pumping cycle.

5) However, when the gradient became great enough so that the flow slowed (Fig. 9, C to D), the left ventricle progressively emptied less completely, and the residual volume resulted in complete left ventricular filling at diastole and interruption of venous inflow at each systole.

6) Whenever the left ventricle was being filled completely at diastole, increasing the mean vascular pressure caused no concomitant

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increase in pump output.

7) As slowing of flow occurred (Fig. 9, C to D) there was gradual increase in a greater than previous volume in the pulmonary circuit at the expense of the systemic circuit. The greater the slowing, the more the volume equilibrium shifted to the simulated pulmonary circuit.

Experiment #4b.   Resistance Near a Pump Inlet.

A metering valve was placed 20 cm. from the left pump's inlet (Fig. 2, #7) across which a pressure gradient could be monitored by way of transducers (Fig. 2, #6). The metering valve was progressively closed. The resulting inflow resistance magnitude is represented by the pressure gradient across the valve. Figure 10 shows the effect of increasing inlet resistance as the two superimposed "CVP" lines (Fig. 10, A to B) become separated (Fig. 10, B to C).
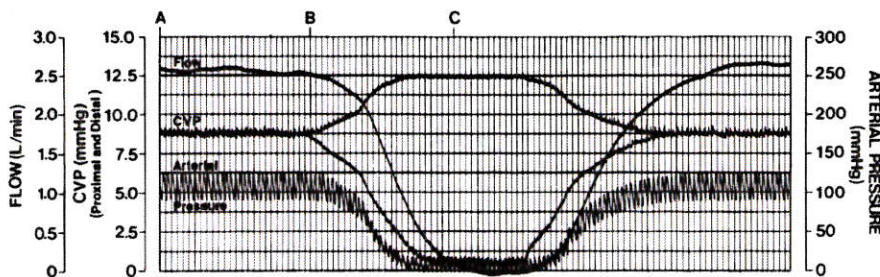

Figure 10.   Relationship of inflow resistance to pump output.
A to B - No resistance at pump inflow.
B to C - Flow progressively decreases as inflow
resistance increases.

Findings:

1) There was a direct inverse correlation between pump output and inflow resistance to the pump (Fig. 10, B to C) over the whole range studied. This is in marked contrast to outflow resistance of the previous experiment (Fig. 9, B to C).

2) As inlet resistance increased there was progressively less filling of the ventricles at diastole. Maximal emptying did occur at each ventricular systole, and venous flow remained uninterrupted at all times.

3) Progressive increase in left pump inflow resistance shifted the fluid volume equilibrium between the pulmonary and systemic circuits toward the pulmonary bed.

4) The pump with the greatest inflow resistance was the major determinant of the output of both.

Discussion: Relationship of Inflow and Outflow Resistances To One
Another and To Pump Output

In that the vascular system is a circle, every point is both up-
stream from a pump inlet and downstream from a pump outlet. There-
fore, every resistance point may offer some inflow and some outflow
impediment to flow. The magnitude of impediment, however, that a
given resistance offers to flow has one other factor; the amount of
vascular compliance upstream from that resistance point.

The less the compliance of the bed upstream from a resistance
point, the greater is the outflow impediment, and the less is the
inflow impediment. Conversely, the more compliant the bed upstream
from a resistance point, the less is the outflow impediment and the
greater is the inflow impediment.

The pump's ability to expend energy to overcome outflow impediment,
and its inability to expend energy to overcome inflow impediment,
makes any generalization about the correlation of pump output to resis-
tance alone, without considering upstream compliance, invalid. It is
only when upstream compliance is added to resistance that factors are
produced that can be correlated with pump output. Therefore, "inflow
impediment" and "outflow impediment" will be used to designate the
interaction of resistances and upstream compliance, which may be a
factor in pump output control in specific situations.

SUMMARY OF MODEL FINDINGS

No single factor correlated with pump output at all times. It was
only when a factor was limiting pump output that it correlated with
flow. Therefore at all times it was the limiting factors that deter-
mined pump output. The limiting factors fall into two mutually exclu-
sive sets of determinants that define two states: "Pump Energy Excess"
and "Pump Energy Failure."

Pump Energy Excess is the state where flow is being limited by factors
other than pump energy. This state was manifested whenever alteration
in pump rate or impeller pressure caused no concomitant change in pump
output. Pump energy excess was always associated with uninterrupted
venous flow to the pumps and submaximal filling of the ventricles at
ventricular diastole. The set of determinants of pump output during
pump energy excess is: mean vascular pressure and inflow impediment
(Fig. 6, Fig. 7-C to D, Fig. 8-B to C, Fig. 9-B to C).

Pump Energy Failure is the state where flow is being limited by pump
energy, in either rate or in strength of impeller pressure, in relation
to outflow impediment. This state was manifested whenever change in
pump rate, impeller pressure or outlet impediment caused a concomitant
change in pump output. Pump energy failure was always associated with
interruption of venous flow at ventricular systole and maximal ventric-
ular filling before the end of ventricular diastole. The set of deter-
minants during pump energy failure is: pump rate and/or impeller
pressure in relation to outflow impediment (Fig. 7-A to B, Fig. 8-A to

B, Fig. 9-C to D, Fig. 10-B to C).

CONCLUSIONS FROM THE MODEL

There are two sets of mutually exclusive determinants of pump output, only one of which can be determining flow at a time. It is the set that is limiting the flow that is the determinant at any time. The power or energy supply available for circulation determines which set at any moment is controlling output:

During Pump Energy Excess:

$$\text{Pump output} = \int \frac{\text{Mean vascular pressure}}{\text{Inflow impediment}} = \int \frac{\begin{array}{c}\text{Fluid volume}\\ \text{system compliance}\end{array}}{\begin{array}{c}\text{resistance}\\ \text{and}\\ \text{upstream compliance}\end{array}}$$

During Pump Energy Failure:

$$\text{Pump output} = \int \frac{\text{Pump rate and/or Impeller pressure}}{\text{Outflow impediment}} = \int \frac{\text{resistance}}{\text{upstream compliance}}$$

PHYSIOLOGIC COROLLARIES TO MODEL FINDINGS

There are corollaries in human physiology for all findings in the model. Furthermore, all circulatory findings in man seem to have a counterpart in one or the other of the two hydraulic states demonstrated in the model: pump energy excess and pump energy failure.

I Myocardial Energy Excess: Pump Energy Excess

This is the hydraulic state where mean vascular pressure and inlet impediment determine the output. Conversely, this is the state where changes in heart rate, strength of myocardial contraction and outlet impediment do not change cardiac output.

A. Cardiac Output Changes Correlate with Mean Vascular Pressure Alterations: (Fig. 6). Mean vascular pressure was first described by Weber more than 100 years ago (17); synonyms include static blood pressure (18), intrinsic blood pressure (19) and mean circulatory filling pressure (20).

During the early stages of hemorrhagic shock, low cardiac output accompanies exsanguination whether or not the heart beats rapidly or strongly (9). Similarly, diminished cardiac output accompanies the decreased blood volume that occurs from dehydration (21) as seen in untreated diabetes mellitus, diabetes insipidus or Addison's disease. In contrast, increase of blood volume by transfusion after hemorrhage increases cardiac output in geometric relation to the volume replaced as long as the vascular tone is unaffected and myocardial function is in excess of the workload (6). Cardiac output above normal levels may develop after over-transfusion or from an abnormally high intravascular

volume such as associated with renal failure (22). High cardiac output also may be associated with high output arterial hypertension when a component of the syndrome is an increased intravascular volume (13).

Cardiac output changes, similar to those in the model, occur where mean vascular pressure is altered by changes in vascular compliance rather than in blood volume. Thus, decrease in cardiac output accompanies any stimulus, reflex or otherwise, which relaxes and thereby increases vascular compliance (23). Vasoactive drugs (e.g., arfonad), metabolic acidosis (24), spinal anesthesia (8), vasovagal syncope (16), and decapitation are all conditions that relax the vascular system and are accompanied by low cardiac output in the absence of any change in blood volume or myocardial energy expenditure. In contrast, high cardiac output results from stimuli that increase vascular tone where there is absence of myocardial energy failure. Circulating catecholamines (after exogenous administration of epinephrine or endogenous secretion by pheochromocytoma) (22), hypoxia, hypercarbia (24), changes in body position and exercise (22), fatigue, emotional tension and anxiety (25) and peripheral nerve stimulation all increase vascular tone, thereby increasing mean vascular pressure which may account for the concomitant increased cardiac output.

B. Cardiac Output Changes Due to Inlet Impediment Alterations: (Fig. 10, B to C). The interplay between inlet impediment and cardiac output during cardiac energy excess is manifested in vena caval obstruction (e.g., inflow occlusion during cardiac surgery, mechanical compression during cardiac massage, or direct encroachment by tumors or thromboses), Valsalva maneuver, venous spasm after administration of vasoactive drugs, and the effects of increased blood viscosity (26). Other examples of inlet impediment correlating with decrease in output include increased atrial impediment from atrial fibrillation and nodal rhythm (22). Moreover, with increased valvular impediment from mitral or tricuspid stenosis, cardiac output decreases in proportion to the severity of the narrowed inlet (27). During rapid tachycardia (7), marked ventricular hypertrophy and pericardial tamponade (21, 28) where filling time is short, chamber size is contracted or distensibility of the heart is restricted, respectively, the output is low.

The increase in cardiac output from arteriovenous fistulae may be explained by the lower inlet impediment to the heart as the fistula bypasses the resistance points that are distal to most of the compliant beds. Furthermore, exercise increased cardiac output where muscle vein pumps tend to overcome the inlet impediment to the heart.

C. Factors Which Have Not Been Shown to Effect Cardiac Output During Myocardial Excess: Increasing Heart Rate (Fig. 7, C to D) and/or Strength of Myocardial (Fig. 8, B-C) Contraction and Decreasing Outflow Impediment (Fig. 9, B to C). The model findings have a clinical counterpart with pacemakers in the range where changes in heart rate do not increase or decrease cardiac output (3, 4). Correlation is also seen when stimuli that strengthen myocardial contractility such as digitalis (10) or which decrease contractility such as propranolol, do not alter

cardiac output except when ventricular energy failure is present (5).
Furthermore, alterations in outflow impediment such as seen in sub-
aortic stenosis, aortic stenosis (15, 22), coarctation of the aorta,
aortic rigidity from atherosclerosis, and arteriolar resistance in
essential hypertension (14) cause no decrease in cardiac output as
long as the heart expends enough energy to effectively eject its con-
tents. The fact that cardiac output in this situation does not change,
does not imply that the output impediment increase in the absence of
failure does not alter the ejection fraction, ejection time, and myo-
cardial metabolism. The very limited secondary elevation in cardiac
output that occasionally occurs during myocardial energy excess by
elevating the strength of contraction or decreasing outlet impediment
may result in a reduction in end-systolic volume which in such cases
may decrease inlet impediment and thus secondarily cause a slight
increase in cardiac output. The resulting change being primarily from
inlet impediment alteration is therefore not an exception to the model
findings.

II Myocardial Energy Failure

The point of myocardial energy failure with the heart is less
abrupt than in the more rigid model; otherwise, the findings have their
counterparts in the model.

A. Low Heart Rate (Fig. 7, A to B). Clinical counterparts to
findings during low pump rate energy failure are seen with the use of
cardiac pacemakers for complete heart block (3, 4). In patients with
heart rates of 20-30 beats/min. the cardiac output is low and ventric-
ular filling is complete before the end of diastole. If a pacemaker
rate progressively increases, cardiac output increases until the ven-
tricles are no longer filled to capacity during diastole. Further
increases in pacemaker rate are then without increase in cardiac out-
put.

B. Myocardial Energy Failure in Overcoming Outflow Impediment
(Fig. 8, A to B and Fig. 9, C to D). Clinical counterparts to the
model are the increase in cardiac output generated when digitalis and
other inotropic drugs (29) are administered to patients during myocar-
dial energy failure. Similarly, administration of propranolol reduces
cardiac output whenever the strength of contraction is decreased to
the point of myocardial energy failure.

Low cardiac output from aortic or mitral insufficiency is from
myocardial energy failure. Inlet or outlet valvular insufficiency may
restrict cardiac output by wasting so much energy on forcing blood
backward that there is myocardial energy failure for forward flow.

Modification of outflow impediment causes cardiac output change
during myocardial energy failure. This is seen as a further corollary
to model findings. Administration of nitroprusside or other arteriolar
relaxing drugs (11) or intra-aortic ballon pumping increases cardiac
output by lowering outflow impediment to a level that low myocardial
energy expenditure is suffcient to cause more adequate emptying of the

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ventricles. Similarly, operative correction of aortic stenosis by valvulotomy or valve replacement (30) derives its benefit by decreasing outflow impediment to a point where myocardial energy is in excess of that used on circulation.

Clinical observations thus correlating with those of the model suggest that myocardial function and its multitude of factors, as they relate to outflow impediment, alter cardiac output only during myocardial energy failure. At all other times myocardial factors provide excess power supply and therefore do not regulate or control cardiac output.

C. Factors That Have Not Been Shown to Effect Cardiac Output During Energy Failure: Mean Vascular Pressure and Inlet Impediment (Experiment 4a, #6). A physiologic corollary is seen during ventricular energy failure where transfusion has no beneficial effect on cardiac output and where surgically creating an arteriovenous fistula for dialysis during heart failure is unaccompanied by an increase in cardiac output.

RESULTING CONCEPT

We may now postulate that:

I. The heart is essentially a non-sucking, continuous inflow, pulsatile outflow pump which therefore, depending upon extracardial factors, may be either active or passive in cardiac output determination.

II. The heart is passive in relation to cardiac output control during pump energy excess, where mean vascular pressure and inlet impediment determine output.

III. The heart is active in determining cardiac output only when it is in pump energy failure, where heart rate and/or strength of myocardial contraction in relation to outflow impediment determine output.

IV. There are two physiologic states which have separate sets of determinants, only one of which can be regulating cardiac output at a time. It is, therefore, always the set that potentially gives the lowest output that limits and therefore becomes the sole determinant of cardiac output.

The two states are:
1) Myocardial energy excess (where mean vascular pressure and inlet impediment to flow determine cardiac output).
2) Myocardial energy failure (where heart rate and/or strength of myocardial contraction and outlet impediment to flow are determinants of cardiac output).

DISCUSSION

The above concept eliminates many of the hard to explain paradoxes

of the past and, therefore, points to reevaluation of the many old concepts. In the new concept, Starling's law becomes an expression of energy conservation during energy excess and has output control implications only during energy failure.

In Poiseuille's law, it is the pressure gardient during myocardial excess and not the cardiac output that changes with alterations in "total vascular" resistance.

Vasomotor tone, with all of its reflex control factors, may alter circulation by changes in mean vascular pressure and inlet imped-iment, and not merely by changing arteriolar resistance.

The new concept identifies such statements as "cardiac output equals stroke volume times heart rate" and "cardiac output equals venous return" as being synonyms of cardiac output rather than being expressions of determinants.

The new insight makes realization that concomitant changes from stimuli are not necessarily responsible for the resulting circulation change. A heart rate change, for example, may not be the factor responsible for a cardiac output alteration resulting from a stimulus.

Pulmonary congestion from high inlet impediment is an example that the clinical designation "congestive heart failure" is not always an expression of myocardial energy failure.

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