A BLOOD PUMP FOR CLOSED-CHEST LEFT VENTRICULAR BYPASS

A Dissertation

Presented to

the Faculty of the Department of Mechanical Engineering

University of Houston

In Partial Fulfillment

of the Requirements for the Degree

Doctor of Philosophy in Mechanical Engineering

---

by

James Dennis Bruner

January, 1969

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ABSTRACT

In the quest for more effective and less traumatic methods of treating heart failure the Baylor University College of Medicine, Department of Physiology, has developed a new method of left ventricular assistance. The method does not require thoracotomy and provides assistance to the failing myocardium. Inherent in this method, however, are some rather severe limitations on the pump and pumping system used in implementing the assist. Most notable among these is a long, small diameter tube (catheter) which is placed in the failing ventricle via a carotid artery in the neck. The catheter is used to withdraw blood from the ventricle for later infusion into the arterial tree. The dimensions of the catheter produce a significant pressure drop which is seen as a vacuum at the pump inlet. This vacuum, with normal pumps, decreases flow rates, increases hemolysis rates and results in other undesirable features of pump performance.

An initial attempt at the design of a pulsatile pump resulted in a collapsible-bag type pump utilizing active, collapsible valves. The pump contained two pumping chambers operated in a push-pull fashion. Evaluation of this pump showed that it was capable of supplying sufficient flow to maintain laboratory dogs with ventricular fibrillation (i.e., total circulatory maintenance) for periods up to four hours. However, hemolysis rates were high and flow pulsations were virtually absent.

A new pump was designed and fabricated, and its operating characteristics were evaluated. This new pump was also of the collapsible-bag type utilizing collapsible valves but contained only one pumping chamber. Included in the design was a vacuum antechamber which, when properly adjusted, provided a reservoir of blood for the pumping chamber which provided a pulsatile outflow. In vitro and in vivo studies showed desirable pulse waveforms and reduced hemolysis.

Based on the results of the studies with the new pump, it appears that with some minor modification it could be of clinical value in left ventricular bypass without thoracotomy.

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NOMENCLATURE

SYMBOL	DEFINITION	UNITS
A	Cross Sectional Area	cm^2
BP	Bag Collapse Pressure	mm Hg
D	Diameter	cm
DC	Duty Cycle	dimensionless
DP	Pump Discharge Pressure	mm Hg
ΔP	Pressure Difference	mm Hg
f	Friction Factor	dimensionless
g	Acceleration due to Gravity	cm/sec^2
h	Height	cm
h_L	Head Loss	mm Hg
ID	Inside Diameter	cm
IV	Inlet Vacuum	mm Hg
L	Length	cm
OD	Outside Diameter	cm
Q	Flow Rate	l/min
r	Radius	cm
R	Pump Rate	c/min
Re	Reynolds Number	dimensionless

NOMENCLATURE (con't.)

SYMBOL	DEFINITION	UNITS
t	Time	sec
V	Average Velocity, Volume	cm/sec, cm^3
VP	Valve Collapse Pressure	mm Hg
μ	Viscosity (poise)	gm/cm-sec
ρ	Mass Density	gm/cm^3

CHAPTER I

INTRODUCTION

It has been over 340 years since William Harvey, in his classic treatise "Exercitato de Motu Cordis et Sanguinis" [47], recognized that the heart is, in essence, a pump which forces blood through the body and, indeed, is the sustainer of human life. Since that day the heart and cardiovascular system have been thoroughly investigated and fundamental ideas changed and revised. One fact, however, has remained intact, that is, that the heart is a pump. Cardiac tissue is muscle and the outstanding characteristic of muscle is its ability to contract; contraction in the heart imparts kinetic and potential energy to the blood in the cardiovascular system. Thus, whatever other functions the heart may have (and other functions are postulated), its primary and major function is that of a pump. That the heart's role is so specific does not in any manner detract from its uniqueness and effectiveness as an organ. The design is exquisite, yet simple and functional.

Mammalian hearts contain two separate pumping systems, each comprised of a two-stage pump. One, constituting the right side, is a low pressure system which drives the blood through the lungs where it takes up oxygen and gives up carbon dioxide; the other, which receives the blood from the lungs, is a high pressure system which drives the oxygenated blood throughout the body. Since these two systems are in series the average flow rate for each must be the same.

Each system makes use of a power chamber (ventricle) in which the output pressure is generated and a supercharger (atrium) which acts as a reservoir and helps prime the power chamber. The pumping action is produced by contraction of the cardiac muscle which ejects the blood from each ventricle simultaneously through leaf-type valves into the vascular system. The inlet to each ventricle is also provided with a valve to prevent regurgitation and the resultant ineffective pumping.

The heart in the human, which is approximately the size of an adult's fist, is given the responsibility of pumping from 3 to 30 liters of blood per minute every minute of every day for the life of an individual; this amounts to a staggering total of nearly 250 million liters in 70 years. If it fails for any reason even for one minute the individual loses consciousness; if it fails for three or more minutes the nervous system is irreversibly damaged and death occurs quickly. It is easy to see why in prehistoric times the heart was held as the most important of organs, possessing the seats of love, hate, courage, cowardice, emotion, virtue, sadness and happiness among many others: all this for a simple pump.

Due to the fact that the heart is essentially a chemical engine providing mechanical energy, it is subject to malfunction, which can be mechanical or biochemical in origin. In actuality the largest killer in the world today is cardiovascular disease of one type or another. Most of these deaths are directly related to cardiac failure, that is, failure of the pump itself. Salisbury [84] has defined heart failure as a condition which is characterized by: (1) abnormally high diastolic pressure within the failing ventricle, (2) excessive venous pressure

(systemic or pulmonary) with adequate cardiac output, or (3) existence of vicious circles that cause progressive hemodynamic and chemical deterioration, and, finally, death. Heart failure may manifest itself in several ways, such as chronic myocardial failure, acute myocardial infarction, congestive heart failure, low cardiac output syndrome following open heart operations, hypovolumetric shock, valvular stenosis or incompetence and cor pulmonale. The actual cause of death may be due to one of many conditions summarized by Salisbury [84].

Classically the treatment for almost all cases of heart failure has been strict bed rest. The reason behind this is that it is felt that the myocardium "rests" when not required to expend large amounts of energy. It is generally believed that the reason behind most cardiac failure is an imbalance between energy supply to the myocardium and energy expenditure required from the heart to maintain the circulation [37]. The feeling is that, in some cases of heart failure, some external method of decreasing the myocardial work load would provide "rest" for the damaged or weakened heart. Therefore, why not introduce another pump into the circuit to relieve the ailing heart of its work load? This is the rationale of assisted circulation.

Heart failure, however, is not the only basis for the use of an "extra" blood pump. Along with myocardial support, blood pumps with auxiliary equipment can be used in many other pathological states including the following [37]: progressive asphyxia, chronic hypoxia, chronic CO_2 retention, metabolic acidosis, renal coma, hepatic coma, exogenous intoxication, heat stroke, accidental

hypothermia, cardiac surgery, total heart replacement and perfusion of isolated organs. With all these possibilities for the use of an artificial heart and associated equipment, it is no wonder that considerable effort has been expended in the development of artificial hearts, oxygenators, dializers and other related equipment. One of the most active of these areas has been the development of heart pumps and, possibly as important, methods of using such a pump. These efforts often occur hand-in-hand to ensure that the pump fits the requirements of the method. Such is the purpose of this investigation, that is, the development of a pump for use with a new method of cardiac assistance.

CHAPTER II

HISTORY OF HEART ASSIST AND BYPASS

Probably the first suggestion that a purely mechanical device could be used in place of a naturally beating heart was made over 150 years ago by LeGallois [61] who proposed that any part of the body could be kept alive by perfusion with arterial blood. Soon thereafter, Blundell [9] showed that purely mechanical devices were not incompatible with the movement of blood. This he did by bleeding a dog from an artery and reinfusing the blood into a vein by means of a syringe. The dog showed no ill effects from this procedure.

Before any work was done in total replacement or assistance of the heart, a long series of experiments was described on the perfusion of single organs and groups of organs. Kay [54] showed in 1827 that the irritability of dying muscle could be restored by infusing blood artificially. Isolated hearts and kidneys were perfused in 1846 and 1849 by Wild [102] and Loebel [66] respectively. The necessity of oxygenating the blood was not recognized until sometime between 1848 and 1858 by Brown-Seguard [11] who experimented with isolated heads and limbs of guillotined criminals. He succeeded in achieving some nervous activity of decapitated heads and showed that deteriorating human muscles could be re-activated by infusion of oxygenated blood.

In 1868 Ludwig and Schmidt [67] constructed a device to provide constant-pressure perfusion of isolated organs. This was a big step over the previous method

of using a syringe. A closed circuit infusion device which included an oxygenator and syringe pump was devised in 1885 by von Frey and Gruber [32] of Germany. Another advance was made by Jacobj in 1890 [52] when he devised a pump and oxygenator system for perfusion of organs. His pump-oxygenator consisted of an appropriately valved single bag collapsed by a cam; the oxygenator was of the bubble type. He later revised his system [53] to include a two-chambered pump, i.e. two bags, alternately collapsed, one of which supplied oxygenated blood to the organ and one which forced venous blood through an oxygenator. The oxygenator in this case was an isolated perfused lung.

Since that time numerous investigators have proposed pumps, oxygenators, and combinations of these to support the circulation of organs outside the human body. Two of the best known of these investigators were Alexis Carrel and the aviator, Charles Lindbergh, who published several papers in the 1930's [14, 15, 62]. Shortly before, in 1928, the first pump designed to replace the heart's function in a living animal was proposed by Dale and Schuster [19]. While this pump was not used to replace the heart in animal circulation, it served as a basis for later design and use of mechanical hearts. Dale and Schuster recognized that the use of lungs as an oxygenator caused fewer problems with perfusion than other oxygenator types. A pump of his own design was used by Gibbs [42] in 1929 to maintain the circulation of cats for up to 3 hours, but he reported no survival.

One field of endeavor which has stimulated the development of blood pumps and oxygenators has been that of cardiac surgery. In 1937 Gibbon [40]

proposed that an artificial heart and lung be used to circulate blood while the patient's own heart was bypassed, thereby providing a bloodless field for cardiac surgery. He later reported the first survival [41] of an animal in which total circulatory maintenance, including oxygenation, was provided.

During the years following World War II interest in mechanical hearts and oxygenators increased immensely. New ideas and concepts in the design of pumps and oxygenators came from all parts of the world. The interest continued to build with increasing success ratios reported in laboratory animals until the first report of the use of a pump on a human was made by Dennis et al. [22] in 1951. The patient unfortunately died. Later in that year Dogliotti [29] reported the first successful partial support of circulation in a human. Then in 1952 Dodrill and associates [26, 27] reported successful temporary substitution of the left ventricle and later temporary substitution of the right ventricle in humans. Finally, Dodrill reported total bypass of a human heart in 1954 [28].

During this period of activity in the development of the artificial heart, the major goal in the back of everyone's mind was either cardiac surgery or individual organ perfusion. At long last, in the middle 1950's the concept of assisted circulation, for reasons other than those mentioned above, began to appear in the literature. The number of methods of assisted circulation are as numerous as the pathological states which might call for the use of a pump, oxygenator or other assist devices. Most of these situations have been mentioned previously in Chapter I. Since the basic problem in this study is that of a pump design, the

discussion will be confined to that of circulatory support rather than respiratory or metabolic support. Galletti and Brecher [36] divide circulatory support into three major areas: (1) veno-arterial pumping, (2) right or left partial bypass, and (3) synchronized ventricular assistance. These are discussed below.

In veno-arterial pumping, blood is drawn from a vein and reinfused under pressure, provided by a pump, into an artery. (See Figure 1.) This method is easily applied and has shown to be effective in certain types of heart failure [25]. It has been widely studied and several investigators have reported on its use [17, 30, 33, 34, 35, 84]. It is important to note that this pumps venous blood into the arterial circulation. Salisbury et al. [83] and Galletti and Brecher [36] state that this method of circulatory assist is particularly useful in right ventricular failure, cor pulmonale and pulmonary congestion.

Ventricular partial bypass, right or left, has also been extensively studied and shows great promise in alleviation of various types of heart failure. Classically, the method involves installation of a pump in parallel with the ventricle in question. That is, blood is drawn from the vein or atrium entering the ventricle and after pumping it is reinfused into the arterial system. (See Figure 2.) By regulating the amount of flow and the method of installation, the flow can be varied from a small percentage of the total blood flow to total ventricular bypass. Until the past few years this method has required the use of extensive surgery for application. Almost all cases required thoracotomy and attachment to the great vessels. Regardless of this shortcoming, the method has shown great possibilities in relief of most types

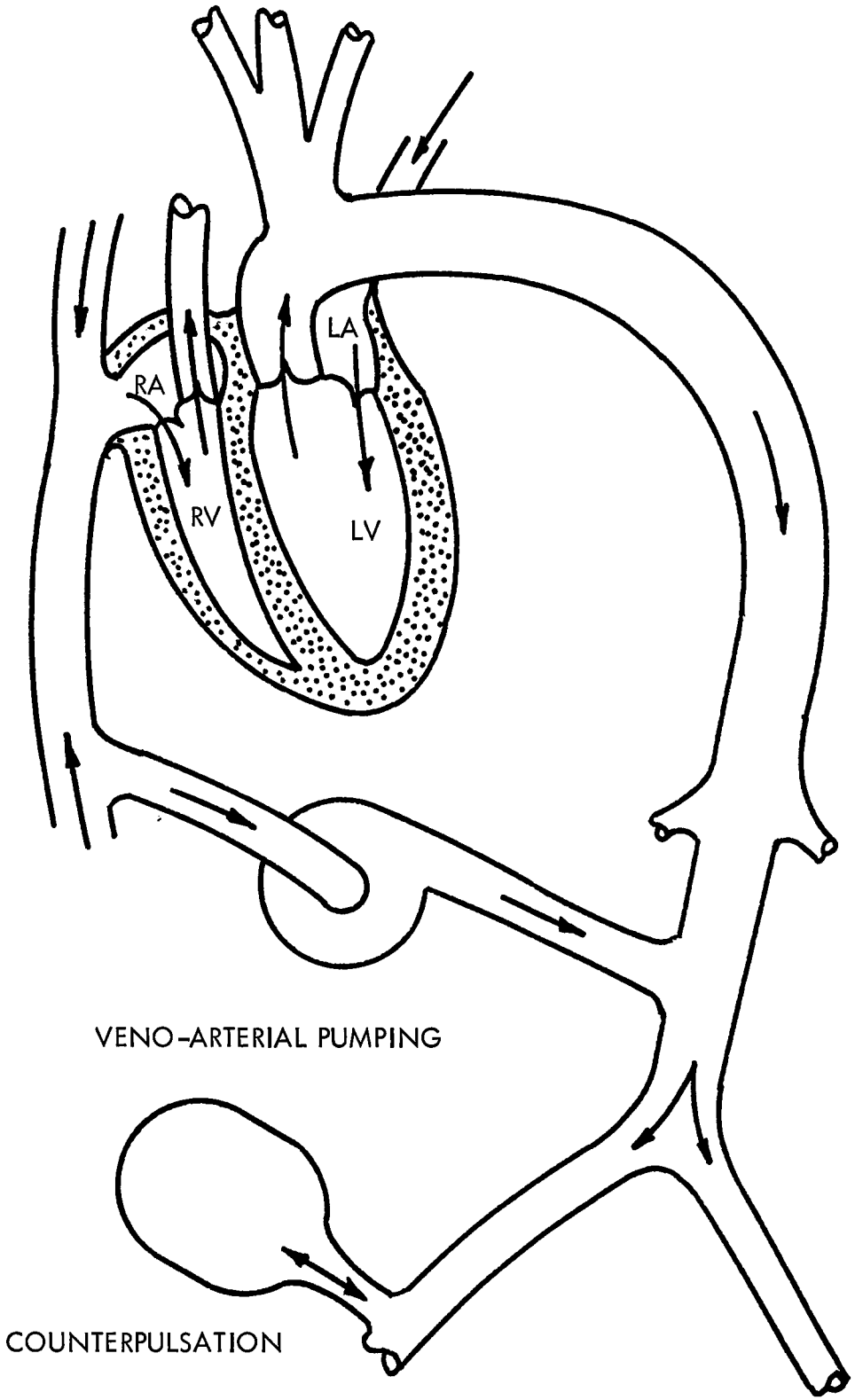


FIGURE 1

METHODS OF ASSISTED CIRCULATION I

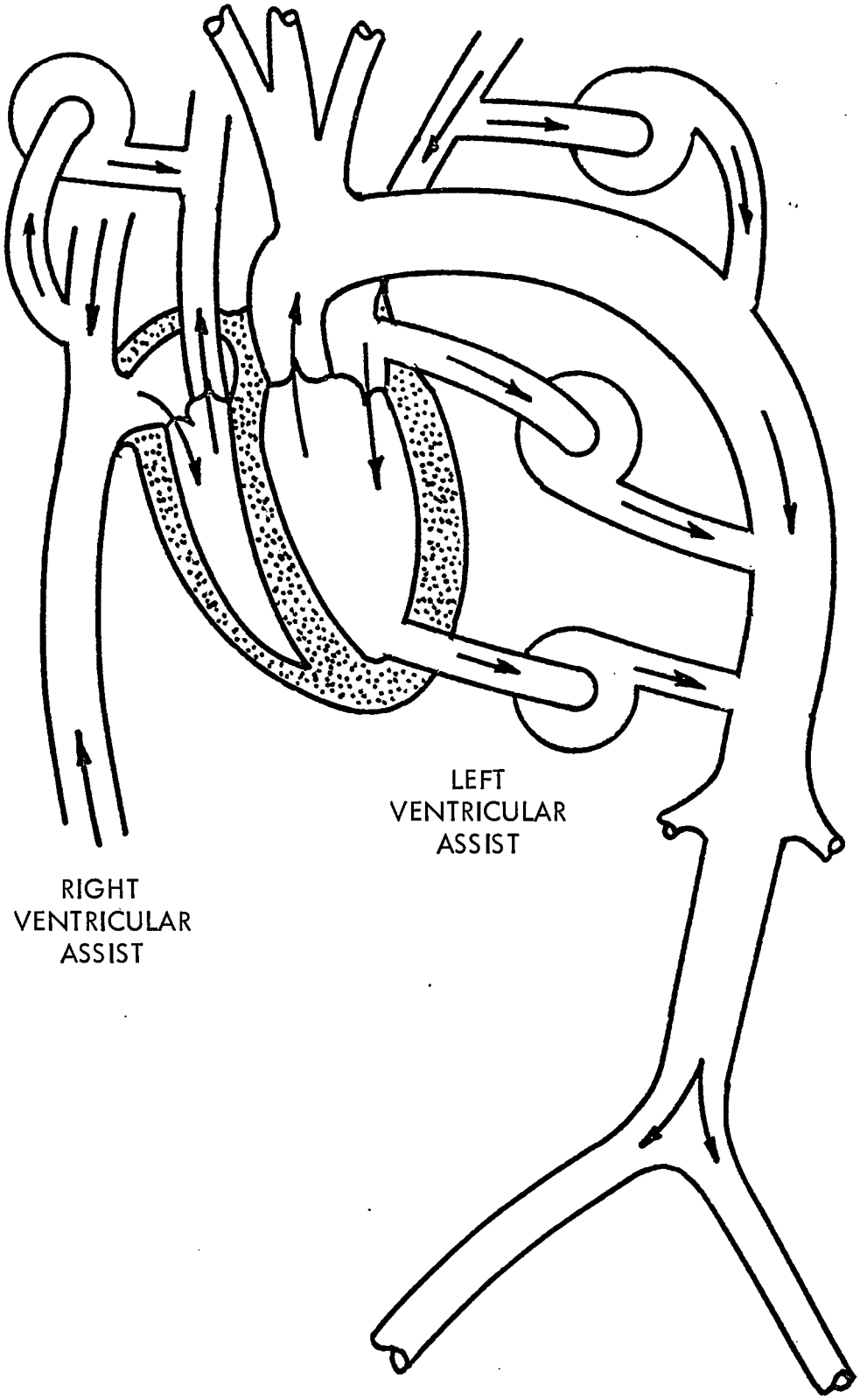


FIGURE 2

METHODS OF ASSISTED CIRCULATION II

of heart failure. Right ventricular assist is of help in patients with right ventricular failure such as congestive heart failure or pulmonary embolism.

According to several investigators [17, 23, 83], left ventricular bypass and assistance is one of the most promising of the circulatory assistance methods for all types of heart failure with the possible exception of cor pulmonale. As previously mentioned, most attempts at left ventricular assistance require thoracotomy and fall into one of three groups as indicated in Figure 2. With these methods the pump inlet is directly connected to either the pulmonary vein, left atrium or left ventricle by suturing of the pump inlet line. Each method has its own group of supporters and each has its own advantages and disadvantages. The first two of these are the most widely practiced and several investigators have reported on their use [63, 90]. More recently Filler et al. [31] have reported on the method in which the pump is attached between the left ventricle and descending aorta. They claim increased effectiveness in left ventricular assist over other methods.

One of the major problems of left ventricular bypass, namely that of thoracotomy, has recently been solved by two groups of investigators. One group, that of Dennis et al. [24], has reported on a method in which a catheter is passed through the jugular vein into the right atrium and thence, via atrial septal puncture, into the left atrium. Another method, that described by Geddes et al. [38] and Schuhmann et al. [86], requires the passage of a catheter through a carotid artery into the left ventricle via the aortic valve.

This method is of particular interest in this investigation and will be discussed later in greater detail (Appendix A). Both of these methods require a minimum of surgery and do not require a thoracotomy, and hence have thus overcome one of the major obstacles.

Many different methods of synchronized ventricular assistance have been exploited. In all of these procedures the method is to reduce artificially the pressure against which the ventricle ejects its blood. This is done by expanding the arterial system during systole and then contracting it during diastole. A brief description of several of these methods is given below.

An auxiliary ventricle, as reported by Nose et al. [73], entails the replacement of part of the aorta by a collapsible bag-type pump. The bag is expanded during systole and collapsed during diastole, thereby reducing the peak systolic pressure of the left ventricle. A similar pump was tried by Brull et al. [12] in which part of the aorta itself was mechanically collapsed.

Birtwell et al. [7] described a method in which left ventricular work is reduced by reduction of intrathoracic pressure in synchrony with systole. This work was further carried out by pulsations in extra-mural pressure in synchrony with the heart beat [8]. These methods have also been studied by Osburn et al. [76].

Still another method involves direct augmentation of ventricular force by means of a sac which encapsulates the failing heart and is expanded and contracted in coordination with normal rhythm. This method is described by Kolobow

et al. [56] and Anstadt [3].

With the exception of the extra- and intramural compression methods, all of the above techniques require thoracotomy to implement. The method of counterpulsation [6, 43, 51, 95, 96] is one approach which does not fall into this category. The counterpulsation device is merely a single-ended pump which is attached to a peripheral artery as shown in Figure 1. During the ejection phase of the cardiac cycle, blood is sucked into the pump and then ejected during diastole. In this way the failing left ventricle is faced by a reduced outflow pressure. The idea is sound and simple and some encouraging results have been reported.

The last method was reported by Laird et al. [60] and involves placing a long thin balloon in the abdominal aorta through the femoral artery. During systole the balloon is collapsed reducing the resistance in the aorta. After cardiac ejection the balloon is filled with air and the pressure within the aorta rises and blood is propelled throughout the arterial tree.

CHAPTER III

REQUIREMENTS OF A BLOOD PUMP

Probably the first artificial blood pump was the simple syringe proposed by LeGallois [61] and used by Blundell [9]. The methods then progressed through the equally simple gravity infusion method for short term perfusion of isolated organs to the collapsible bag pump of Jacobj [52, 53]. Since that time almost every conceivable type of pump has been proposed for use with blood. Some of the most successful basic designs have been the diaphragm pump, fashioned after Dale and Schuster's [19] model, and the roller type as developed by DeBakey [20], Henry and Jourelet [48] and others. Other popular types have been finger pumps, such as the Sigmamotor Pump [36, p. 54] and the collapsible-bag or ventricular-type as used by several investigators [10, 49, 57, 97, 98]. Use of each type is accompanied by its own advantages and disadvantages.

Since there have been so many designs for blood pumps, one naturally wonders about the important factors that determine an acceptable device. These factors can be divided into two categories: those which are truly minimum requirements and those which may be considered desirable but not absolutely necessary. The necessary factors are:

1. It must be able to maintain an adequate flow rate and pressure head.
2. Trauma to the blood must be minimized.

3. Formation of blood clots must be delayed as long as possible.

4. If the pump is to be used clinically, provision must be made for operation in case of total power failure. High reliability of components is also necessary in this case.

5. Sterilization must be possible.

These factors are discussed in greater detail later, along with certain of the following desirable features:

1. Ease of disassembly, cleaning and sterilization. Whenever possible, components in contact with the blood should be disposable.

2. Flow rates should be variable and controllable by the operator. Included here should be a method of determining flow rate at any time during operation.

3. Minimum priming volume.

4. The necessity of pulsatile flow is debatable.

5. Cost factors are important.

Flow Rate. The requirement that adequate flow rate and pressure be maintained simply means that sufficient oxygenated blood must perfuse the body tissues. The flow rate for adequate tissue perfusion is a function of many variables but is usually thought to have a minimum of 50 ml/min/kg [68, 79, 81]. One source [75] states that a flow rate as low as 20 ml/min/kg is sufficient provided that venous pressure remains high.

Blood Trauma. It is generally accepted that all blood pumps, including

the biological heart, cause damage to the fluid being pumped. The degree of damage is a function of many variables of the pump and its circuit. Most of the damage appears as the destruction of red blood cells, with lesser damage to the other cellular and protein constituents of the blood. Immediate red blood cell destruction appears as the release of free hemoglobin into the blood plasma (hemolysis) and is the usual method of quantifying blood trauma. This is not a complete picture of the red blood cell damage as there is evidence that pumping may reduce the average life span of the red blood cell as suggested by Bernstein et al. [4], Kusserow et al. [59], and Shea et al. [88]. Plasma hemoglobin level, however, is presently the best immediate indicator of blood trauma.

One of the first reports to show that artificial pumps might cause damage to the blood was due to Dennis et al. [21] in 1951 in which failure of their procedure was attributed to blood trauma. Since that time many investigators have studied the situations which contribute to hemolysis. Some of the more important factors are the following:

1. Blood Velocity: Bernstein et al. [4] have shown that high flow velocity is one of the more important factors contributing to hemolysis. They stated that, at average velocities exceeding 2000 cm/sec, hemolysis rates increased. This seems to be more important than the occurence of turbulence which probably occurs below this velocity.
2. Material: It has been shown [93] that certain materials are less

hemolytic to blood than others. The desirable factors are surface smoothness and non-wettability. Some of the most widely used materials are silicone-based materials, Dacron and Velour-lined materials [1, 65] which develop their own endothelial linings and the Tygon-type plastic tubings.

3. Pressures and Pressure Fluctuations: Studies [4] have shown that steady positive and pulsating pressures as high as 3 atmospheres are not damaging to blood. Pressures below atmospheric, however, appear to be more hemolytic [4, 103], particularly if the pressure falls 300 mm Hg below atmospheric pressure. If the vacuum is associated with aspiration of air, hemolysis increases rapidly [71].

4. Occlusiveness: Whenever any two surfaces come into contact, red blood cells are trapped between and damage results. It is believed [5] that this phenomenon is one of the major factors contributing to hemolysis in roller pumps and others in which such contact is made.

5. Blood Condition: It is known that different bloods have different red blood cell fragilities which can be the result of many siutations. One of the most important is blood lipid level. Experimental studies [4] have shown that the blood of animals which have been fasted for 12 hours or more is much less subject to trauma when compared to the blood of animals which have recently eaten.

Several methods describing hemolysis have been proposed. Actual hemolysis levels are usually given as milligrams (mg) of free hemoglobin per 100 milliliters (ml) of plasma, also termed milligrams percent. To evaluate pumps in terms of hemolysis, the usual term has been the index of hemolysis (IH) [2], which is the amount of hemoglobin released per 100 ml of blood pumped and is calculated from the equation

$$IH = \frac{(100-Hct)}{100} \frac{(\text{mg free hemoglobin})}{Qt/V}$$

where Hct = blood hematocrit,
Q = flow rate, l/min,
V = system volume, ml,
t = duration of test in min.

Values of IH for several pumps are presented in [4]. Roller- and ventricle-type pumps seem to have the lowest values, in the range of 0.1. However, different investigators report wide variations for evaluations of the same pump. One attempt to improve this evaluation has been proposal of the use of a traumatic index (TI) by Koller and Hawrylenko [55]. The TI is defined as the amount of free hemoglobin released by the pump in performing 100 strokes. The major advantage to this work is a parallel standardized test to normalize the results to different blood fragilities. The work in the Baylor Laboratories has shown that this is an important factor. It does, however, complicate the test procedure.

Blood Clotting. Another problem with artificial circulation devices is thrombus formation within the circuit. Almost all pumping procedures require

the administration of anticoagulants such as heparin. This is not the only solution, however, and factors contributing to clot formation should be eliminated. They are:

1. Surface Smoothness: Rough surfaces favor clot formation [64]. Silastic- and Tygon-type tubings have smooth surfaces and are quite anti-thrombogenic. Coating materials with silicone compounds such as Siliclad compound (Clay-Adams, Inc.) helps reduce surface roughness and minimize the problem. Another solution to the problem has been the coating of surfaces with a heparin compound which helps delay clotting [45, 50]. There is come evidence, though, that this procedure increases hemolysis [5].
2. Probably the most important single factor is that of the presence of regions of stagnation within the system. A considerable amount of clotting is a result of blood stagnation. Thus, any device which pumps blood should be constructed with as smooth an internal bore as possible to avoid stagnation regions.
3. Junction points between materials are most dangerous points for clot formation. Nose et al. [74] state that these areas are particularly bad when the junction is between a smooth and a rough or mesh surface.

Pulsatile Flow. Volumes have been written on the necessity for or the

unimportance of pulsatile flow. The list of investigators is long and the results are inconclusive. Studies have centered around lymph formation and flow [77, 78], kidney function [39], peripheral vascular resistance [69, 70, 99], electrolyte and ion balance [44, 87], peripheral blood pooling [72], and central nervous system function [100]. Whereas one school believes that pulsatile flow enhances the effect of the flowing blood, the other school seeks to show that the pulsations have negligible effect. The question always asked is "Why is pulsatile flow necessary, or are the deficiencies and problems encountered merely due to the limitations of a biological pump?" Let the controversy be terminated in this study by asking the question, "Why NOT pulsatile flow?" We know that this type is physiologically acceptable to the body (and indeed it evolved with it); and if there are no reasons, and there appear not to be, that a properly designed pulsatile pump should not be used, why not use one?

CHAPTER IV

THE DESIGN OF A BLOOD PUMP

Preliminary Considerations

The preliminary work in the Baylor Laboratories on the development of the Geddes-Schuhmann method of left ventricular bypass was done with a roller-type pump. (This is discussed in Appendix A.) This type of pump was used primarily because one was available. Roller pumps do, however, have many desirable features, including ease of operation, quietness, simple construction, ease of cleaning and sterilizing, disposability of components in contact with blood, development of low hemolysis rates, and ease of adjustment. Roller pumps are widely known and their performance has been proved clinically. Roller pumps do have, however, some features which prompted consideration of other types of pumps. These features include:

1. Non-pulsatile flow: Pulsations available from a roller-type pump are small at best. Any compliance, such as air chambers and long connecting lines in the system, reduces what little pulse is available. Since the question of the necessity of pulsatile flow has not been adequately answered, it was felt that pulsatile-type flow should be adopted.
2. High vacuum inlet: Several problems are encountered as a result of the small diameter inlet catheter, which is a necessity of this

method.

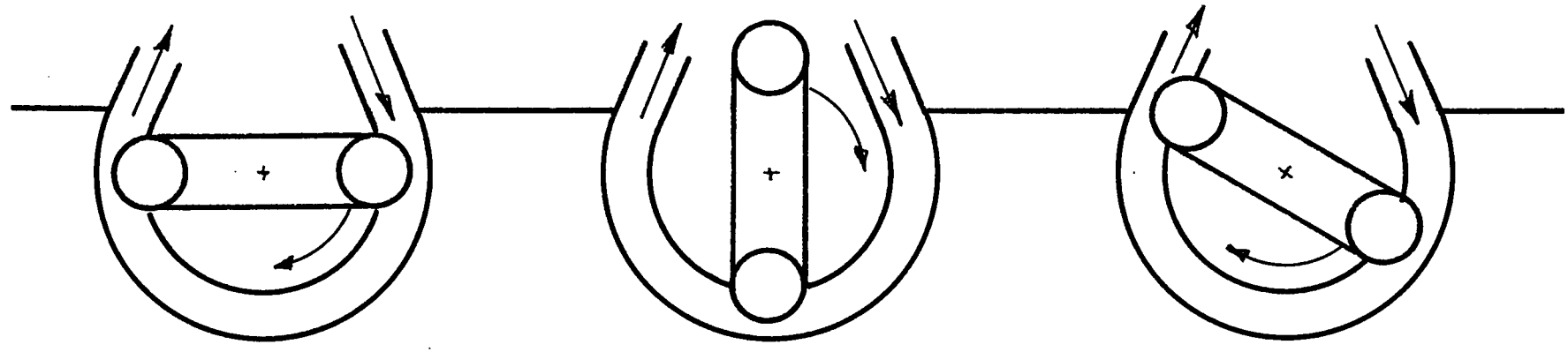
a) Vacuum difficult to control: The only method available to control the inlet vacuum is the pump rate, and this is only an indirect control at that, since the vacuum is really a function of the stiffness of the pump tube. More rapid emptying (i.e., faster rate) is then the only method of controlling this variable.

b) Low flow rate: The restricted inlet often causes a roller pump tube to remain nearly collapsed. In fact, at an operating inlet vacuum of 350 mm Hg, a standard pump tube collapses to approximately 2/3 of its volume at atmospheric pressure. At this inlet vacuum, the maximum flow rate for this pump is 1.2 liters per minute, a near-borderline amount.

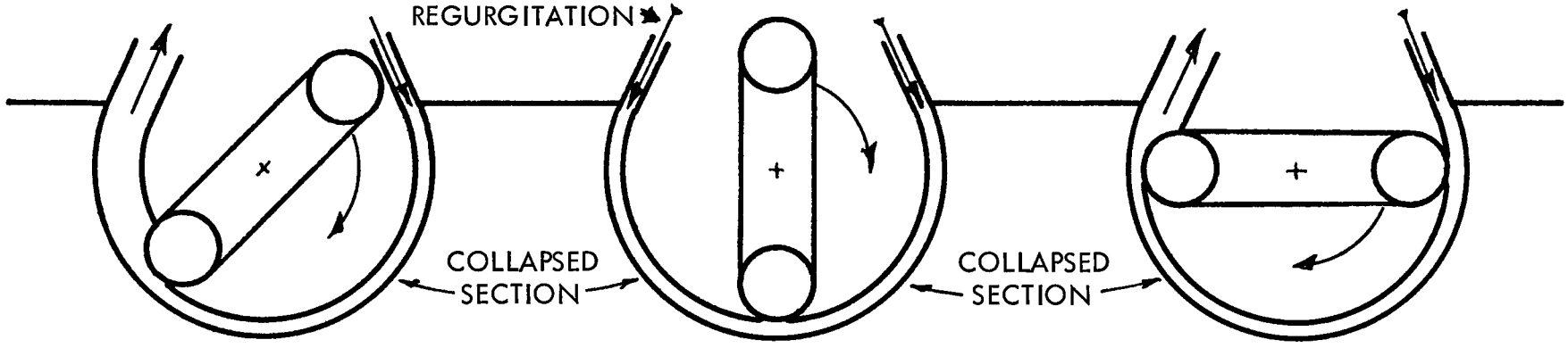
c) Regurgitation: The restricted inlet causes a rather unusual problem with the roller pump which arises due to the lack of valves in the flow lines. Valves, for normal operation of roller pumps, are, of course, not necessary due to the occlusive nature of the rollers. During most of the pumping cycle, only one of the rollers is in contact with the tube. The area of the tube behind the roller is expanding, due to its stiffness, to fill with blood while the roller is

pushing the blood in front out into the systemic circulation. With a restricted inlet, the tube behind the roller often does not have sufficient stiffness to expand fully in the time of one half cycle. As a result, when one roller reaches the end of its travel, it opens the not-yet-fully expanded (i.e., collapsed) tube to the systemic arteries. The result is regurgitation, or the drawing of blood from the arterial system into the pump. (See Figure 3.) This phenomenon is detrimental because it causes alternating collapse and expansion of the arterial tree. The problem could be alleviated by the installation of a check valve in the discharge line.

3. Non-synchronous nature: Another characteristic of roller pumps is that since they are powered by electric motors, they cannot be synchronized with the normally beating heart. This is not a problem in clinical heart bypass for surgery since the heart is not functional. When trying to reduce the work load of a failing heart, however, it can be quite important. It will be recalled that one goal of assisted circulation is to reduce the pressure against which the heart must pump. In left ventricular bypass using non-pulsatile flow, the beating heart must pump against the systemic pressure maintained by the pump. If the pump has



NORMAL OPERATION



RESTRICTED INLET OPERATION

FIGURE 3

EFFECTS OF INLET RESTRICTION ON ROLLER PUMPS

the capability of being synchronized with the beating heart, infusion can be made following cardiac ejection. As a result, the beating heart must pump against only diastolic pressure and not mean pressure. Another advantage of synchronization is the following. It is accepted that minimum coronary blood flow resistance occurs during myocardial relaxation (diastole) [82]. Since substantial coronary blood flow is necessary for a recuperating heart, it would be desirable to have maximum blood pressure occur during this period of low flow resistance. Such is the characteristic of a properly synchronized pulsatile pump.

The one factor which is a function of the system and remains invariant regardless of the pump used is the inlet catheter and the resulting restriction to flow. It is desirable to know the pressure drop through this catheter, the diameter of which is usually dictated by the vessel to which it is connected. A theoretical study for a typical catheter has been made, and the details of calculation are shown in Appendix B. It was found, for example, that to maintain a blood flow rate of 2 liters per minute at normal body temperature and a blood hematocrit of 45 requires a net driving pressure of 270 mm Hg. This flow rate yields a flow velocity of 472 cm/sec (well below the hemolysis level of 2,000 cm/sec) and a Reynolds Number of 6840 which indicates the existence of turbulent flow.

Description of First Pump

In order to develop a pulsatile pump for use with the closed-chest method

of left ventricular bypass, L. A. Geddes and A. G. Moore of Baylor University College of Medicine proposed a single collapsible bag-type pump using active valves. The bag and valves were actively collapsed and expanded by alternating application of air pressure and vacuum to the chambers surrounding the bag and valves. Both the bag and valves were enclosed in plastic (Lucite) chambers to contain the air-pressure variations. A cross section of this pumping chamber is shown in Figure 4. Switching of air and vacuum to the bag and valves was accomplished by small electrically operated solenoid valves of which three were required: one for the bag and one for each valve. Pressure regulators installed in the pressure lines allowed independent control of the pressures applied to the valves and bags. Surge tanks in the lines which were supplied by separate electrically powered vane-type pumps reduced transient variations in pressure and vacuum. Timing and synchronization were controlled through a small electrical control unit; rates from 33 to 250 cycles per minute were available. The pump control unit also had the capability of varying the duty cycle (i.e., percent of the cycle devoted to fluid ejection) from 33% to 50%.

All components including the bags and valves were fabricated in the Baylor Laboratories. The bags and valves were made from a heat-curing silastic rubber (Dow Corning 372), which contains a Dacron weave for strength. In the uncured state, this material can be shaped and molded around almost any form. It clings to itself and becomes bonded when heat cured. Bags were formed around a plaster mold by the overlapping of six sections, whereas valves required

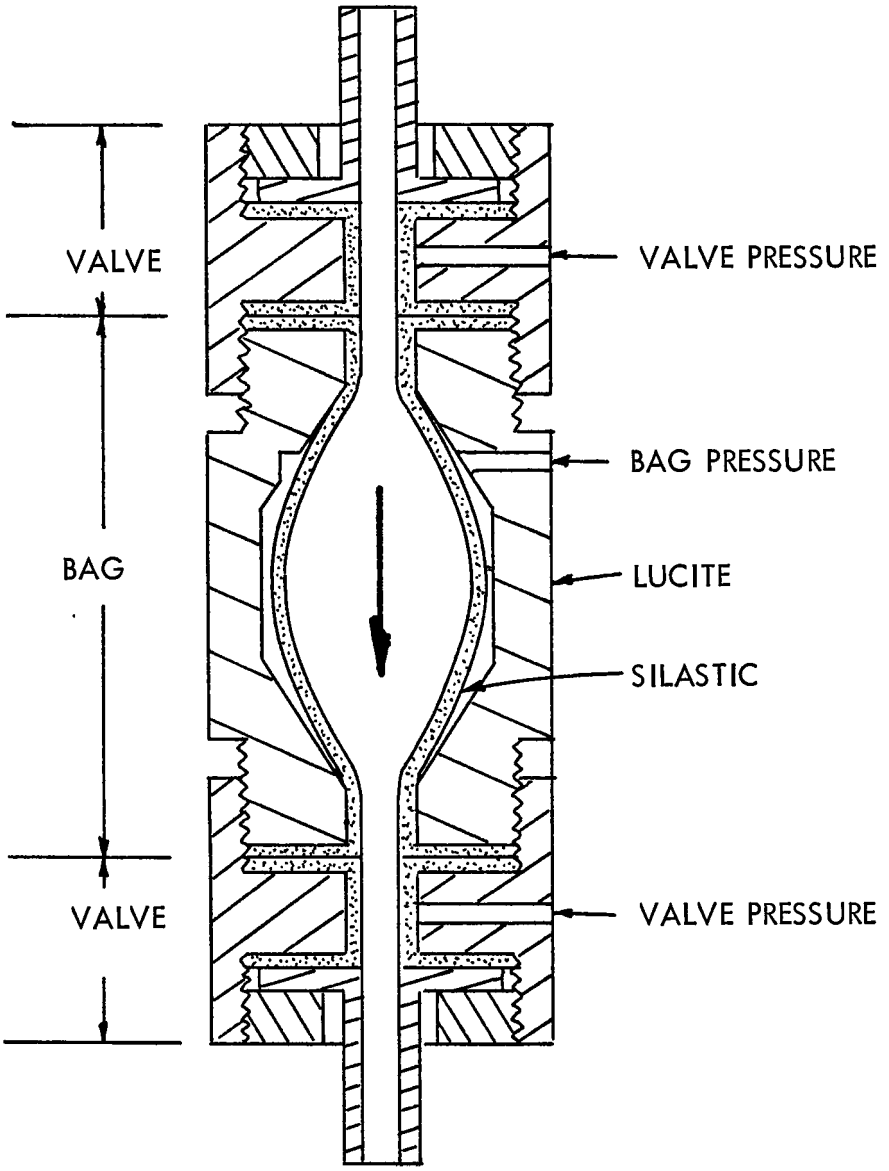


FIGURE 4

CROSS SECTION OF OLD PUMPING CHAMBER

only a single piece overlapped once. All bags and valves were fitted with end flanges, which were used to provide a seal when squeezed between the plastic sections as shown in the figure. Each valve was approximately 1/2 inch in diameter by 1 1/2 inches long. The bags were 4 inches long and 2 1/2 inches in diameter at the maximum and had a volume of approximately 110 ml. The Silastic material, as well as being easily formed, is non-hemolytic and relatively antithrombogenic.

This investigator entered the bypass program after fabrication of the first model of this device, and he was given the project of developing the pump for use with the described method of left ventricular bypass. At this time, the first model was capable of delivering over 3 liters per minute through 3/8 inch diameter inlet and outlet tubes. The only time that this pump was tried on an animal, using the Geddes-Schuhmann method of left ventricular bypass with fibrillating ventricles, the test had to be terminated after 15 minutes of pumping due to insufficient blood flow. Maximum flow was achieved with the 33% duty cycle. The main problem with the pump was insufficient filling of the bag. Based on the results of this test, it was decided to modify the pump design. The new design included the following:

1. Parallel and alternately actuated pumping chambers: Each chamber draws from a common inlet line attached to the intraventricular catheter and pumps into a common collecting chamber. Two outlets of the collecting chamber are provided: one for infusion at

the femoral artery and one for infusion at the carotid artery, as mentioned in Appendix A. Each pumping chamber is identical in design and construction to the original pumping chamber.

2. On-line Bourdon pressure gages to monitor pump inlet and discharge pressures: These two variables are important to the operation of the pump, and their addition has been valuable. Each gage is supplied with convenient electrical readout for the Physiograph (E & M Instrument Co., Inc.) if desired. Measurement of pressure is made at the following points: inlet vacuum at the coupling between the small diameter ventricular catheter and the larger diameter pump inlet line, and discharge pressure at the collecting chamber.

3. New air- and vacuum-control valves: Due to the fact that the two pumping chambers required an additional control valve, it was decided to replace them all. The new valves (Humphrey Products) have larger ports, open and close faster, and allow the passage of air with less flow resistance. These valves are actuated by compressed air at 30 psig and triggered electrically. This reduces the electrical power requirements without sacrificing actuation time (approximately 20 ms.). Again air pressure is supplied by electrical pumps or any other convenient air supply such as bottles of compressed air. A diagram of the pneumatic system is

shown in Figure 5.

4. This particular model could not have the capability of varying the duty cycle since any duty cycle other than 50% would cause both chambers to be filling or emptying at the same time. Also, no provision for valve lead or lag is made. The electrical triggering signals to both the bag and valves are made at the same time. Competency of the valves is assured, however, due to the much larger volume of the bag chamber as compared with the valve chamber. Visual observation of the operating pump shows almost no regurgitation.

5. A new electrical control unit was later fabricated which increased the frequency range. This unit contains a redundant output system in case of failure of one system. A pump flow rate transducer was added to facilitate calculation of pump output. This device consists of two electrical contacts in the blood stream. Pump output is calculated by the saline-conductivity method of Smith [91, 92]. Experiments have shown a calibration factor for this cell of 50 ohms equal to 0.78 grams per liter of saline. This transducer has shown to be unusually stable and sensitive.

Evaluation of First Pump

The evaluation of the pump was carried out in three phases. The first

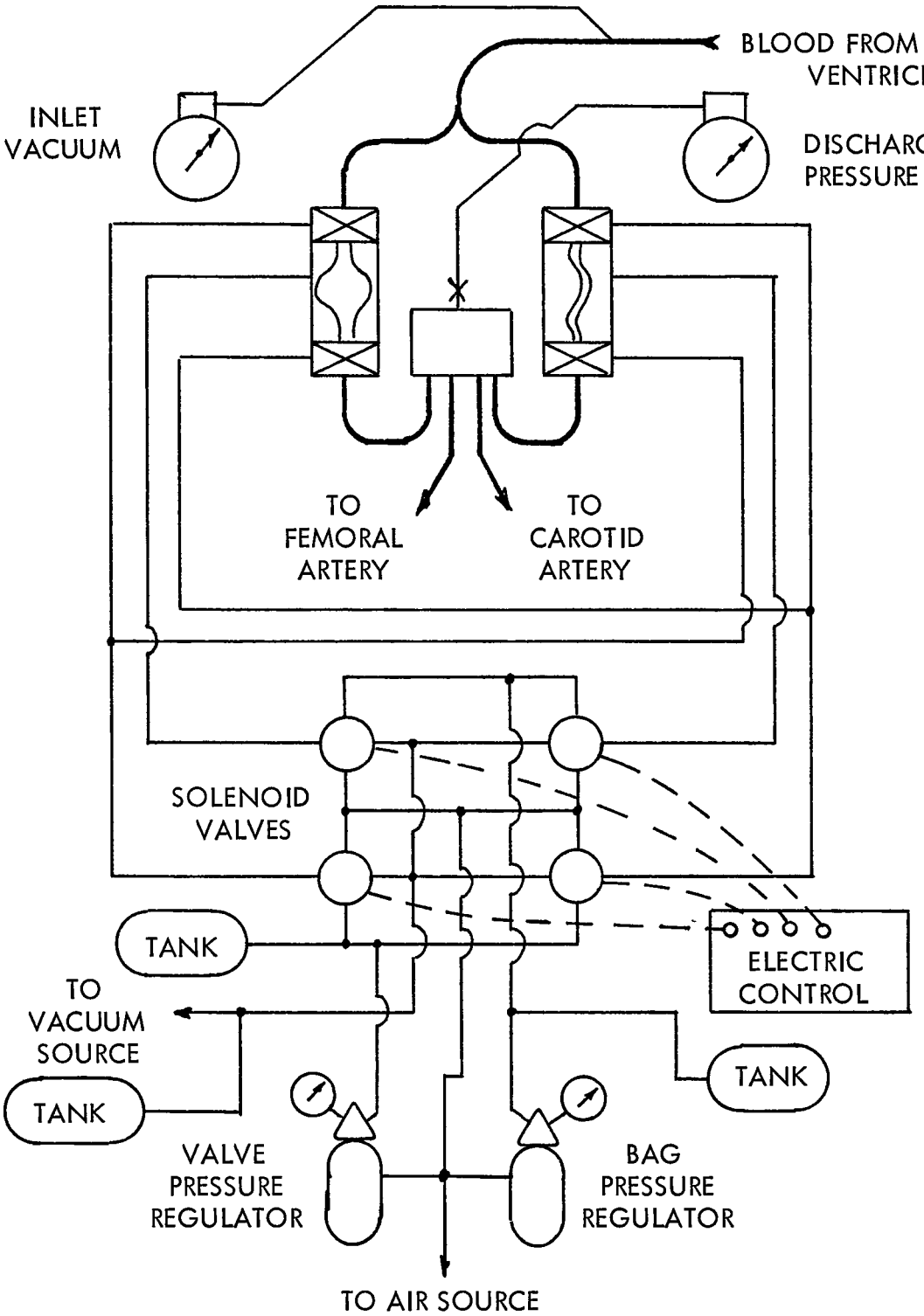


FIGURE 5

PARALLEL PUMP BLOCK DIAGRAM

phase was concerned with flow studies in which water was the working fluid. The second phase was in vitro hemolysis studies to determine blood trauma caused by the pump and system. The final phase was a series of in vivo experiments in which the pump was used with living animals.

Flow Studies.

Valve Competency Test: This test was designed to measure the effectiveness of the collapsible-type valves. A standard valve made of Silastic sheet 0.020 inches thick was used in the apparatus shown in Figure 6. Valve pressure and fluid (water) drive pressure were controlled by pressure regulators. Plots of flow rate through the valve versus fluid drive pressure were constructed for constant values of valve-collapse pressure and are shown in Figure 7. The parameter of interest turns out to be the difference between the valve collapse pressure and fluid drive pressure, as shown in Figure 8. As expected, these data show that the competency of the valves increased as the pressure difference increased. Thus, for maximum competency, the valves should be operated at as high a pressure as possible.

Pump-Flow Studies: Using water as a working fluid allowed a series of studies to be performed to determine the operating characteristics of the pump in terms of its operating parameters. The operating parameters were of two types: those which were characteristic of the pump, such as pump rate (R), bag collapse pressure (BP), valve collapse pressure (VP) and valve and bag vacuum; and those

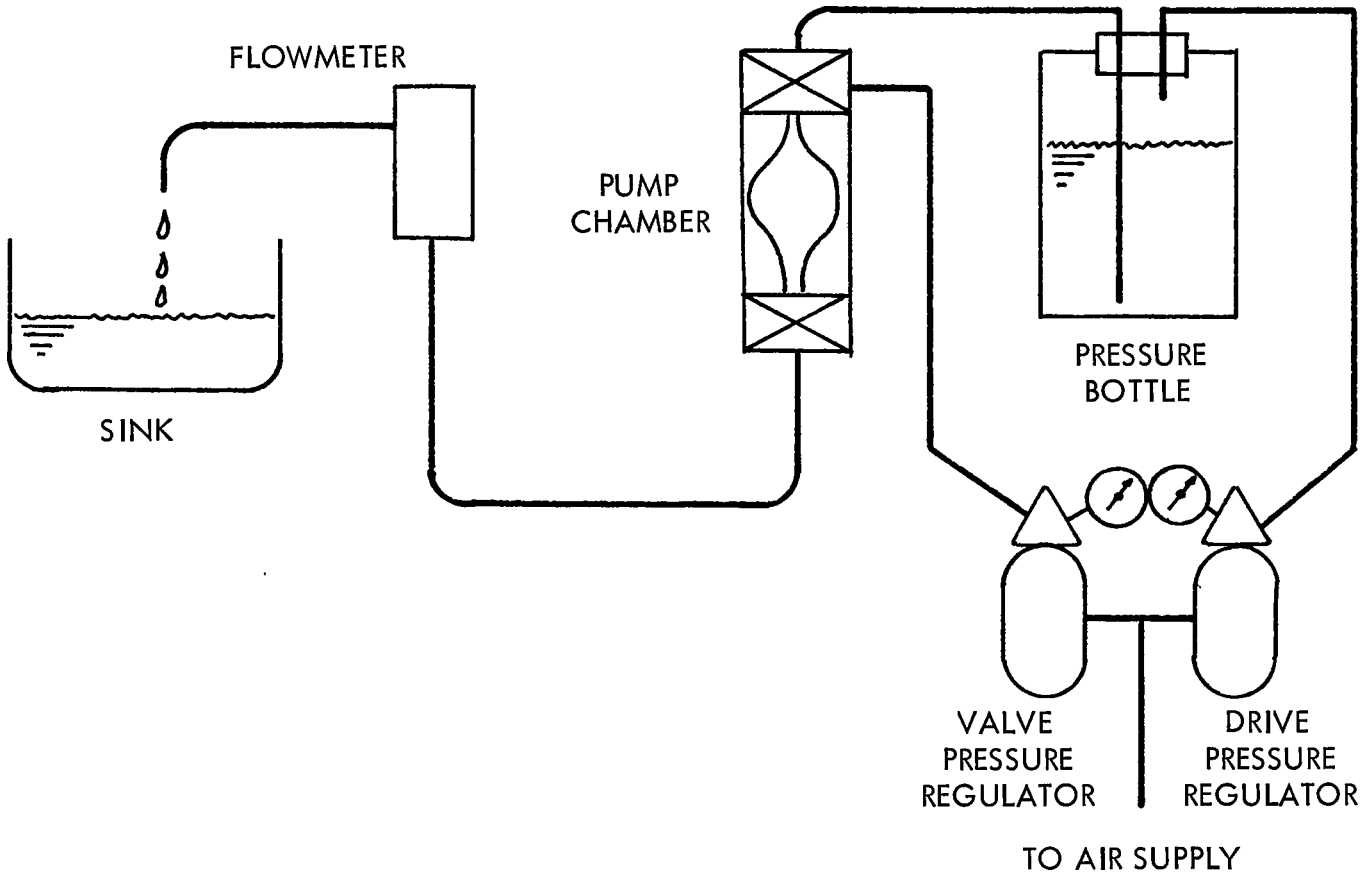


FIGURE 6

VALVE COMPETENCY TEST APPARATUS

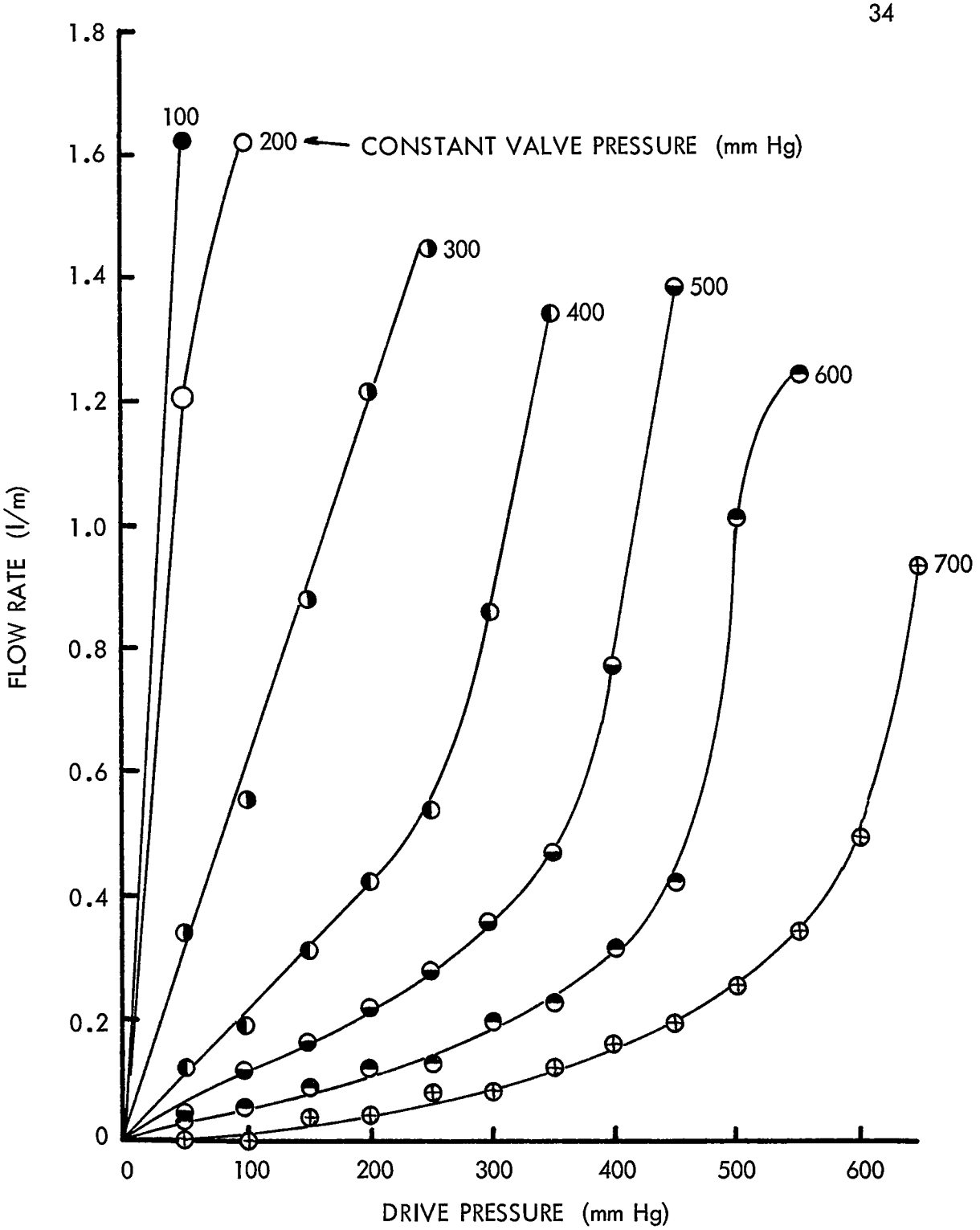


FIGURE 7

VALVE LEAKAGE VERSUS DRIVE PRESSURE

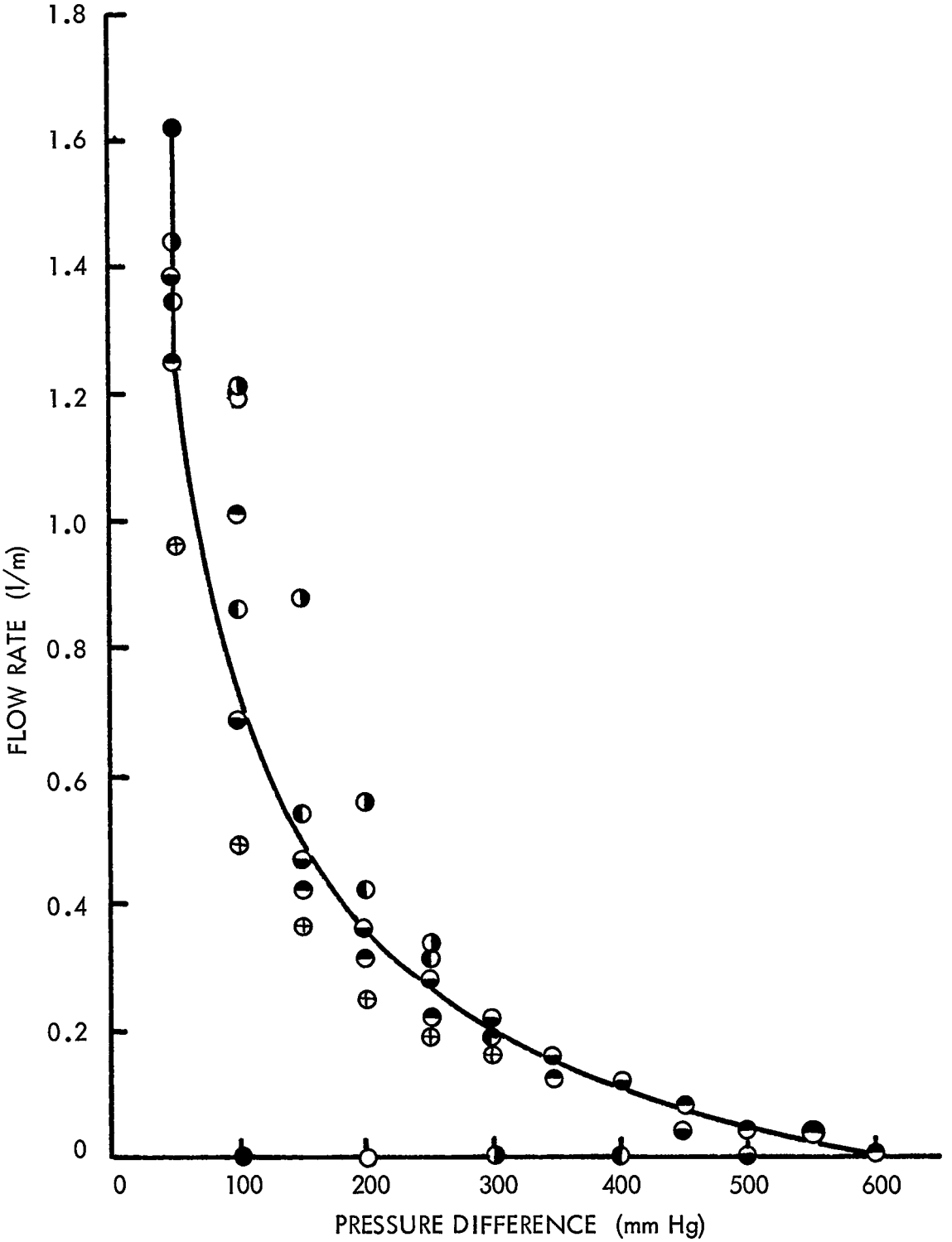


FIGURE 8

VALVE LEAKAGE VERSUS PRESSURE DIFFERENCE (VALVE-DRIVE)

which were mainly characteristics imposed on the pump by the system into which it pumps, including pump inlet vacuum (IV) and pump discharge pressure (DP). All of the above, with the exception of the valve and bag vacuum, were considered variable and were varied in the ranges which occurred during in vivo experiments with the pump. The bag and valve vacuum was not varied as it is not usually varied during the course of in vivo experiments. The above mentioned parameters were varied in the ranges listed below.

TABLE I

OLD PUMP FLOW STUDY PARAMETERS

Parameter	Range Tested
Pump Rate (R)	40, 50, 60, 70, 90, 120 (cycles per minute)
Valve Collapse Pressure (VP)	700 mm Hg
Bag Collapse Pressure (BP)	50, 100, 200, 300 mm Hg
Pump Inlet Vacuum (IV)	0, 100, 200, 300, 400 mm Hg vacuum
Pump Discharge Pressure (DP)	0, 100, 200, 300 mm Hg*
Bag and Valve Vacuum	500 mm Hg vacuum

* and 50 mm Hg less than bag pressure.

In all cases the dependent variable of interest was the flow rate (Q) in liters per minute. The results from these tests are shown graphically in Figures 9, 10, 11, and 12. Meaningful presentation of data with as many variables as studied here is difficult at best. About the only method available is three-dimensional plots representing surfaces. Plotting several surfaces on one set of coordinates handles most of the variables. Whereas these plots are useful for forming opinions on general pump performance, quantitative information is

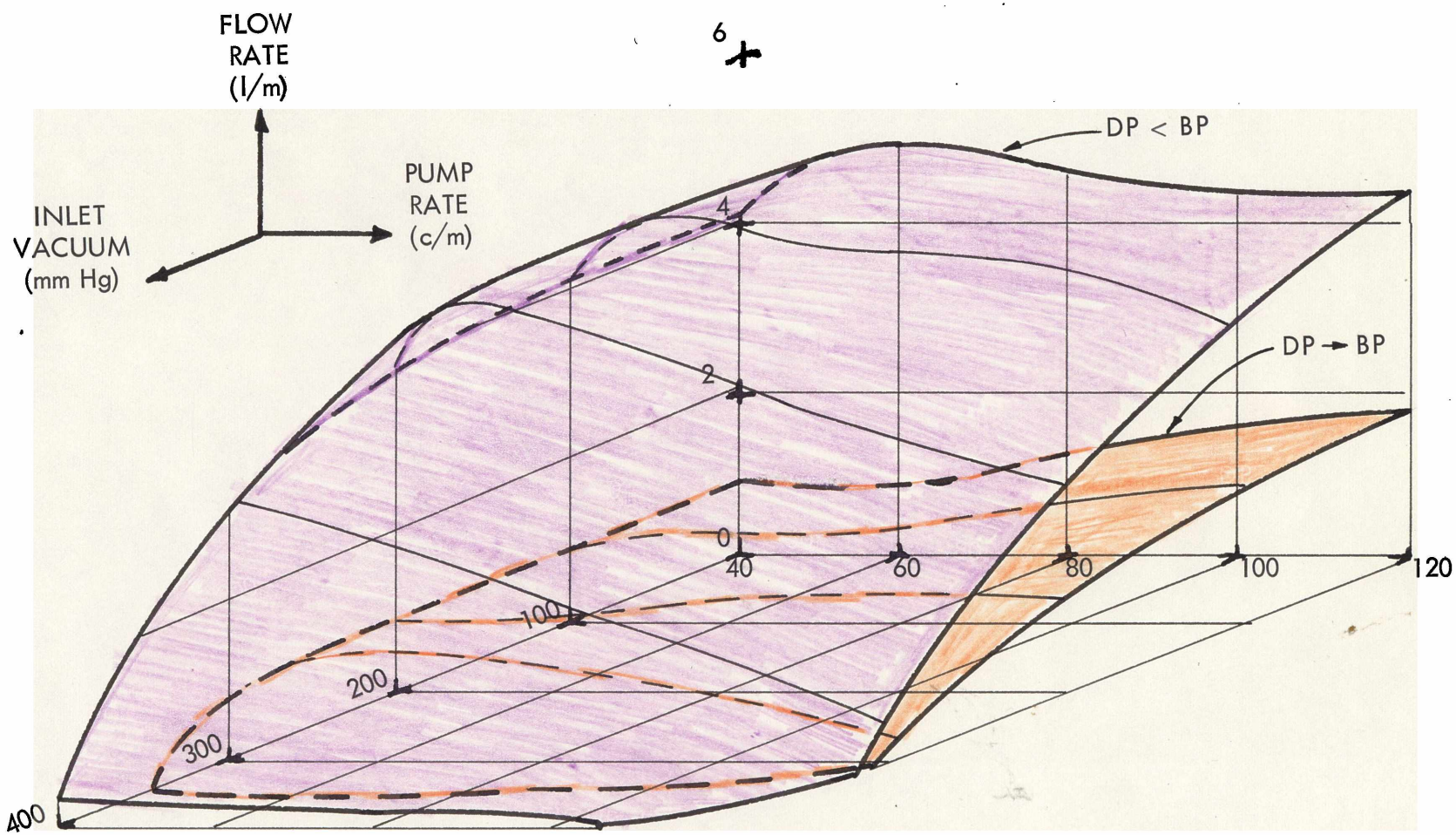


FIGURE 9

FLOW RATE, PUMP RATE, INLET VACUUM SURFACES, BP = 100 mm Hg

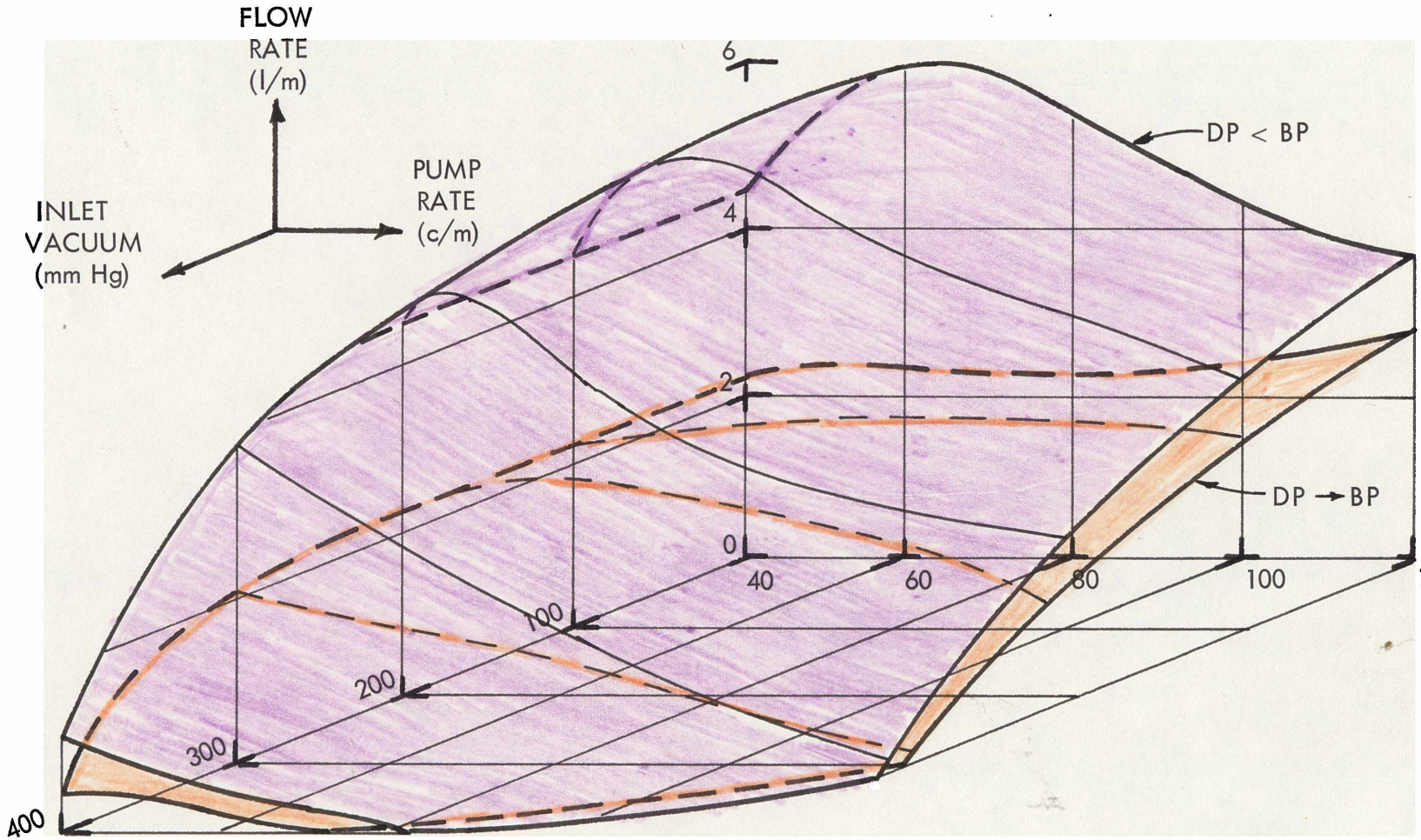


FIGURE 10

FLOW RATE, PUMP RATE, INLET VACUUM SURFACES, BP = 200 mm Hg

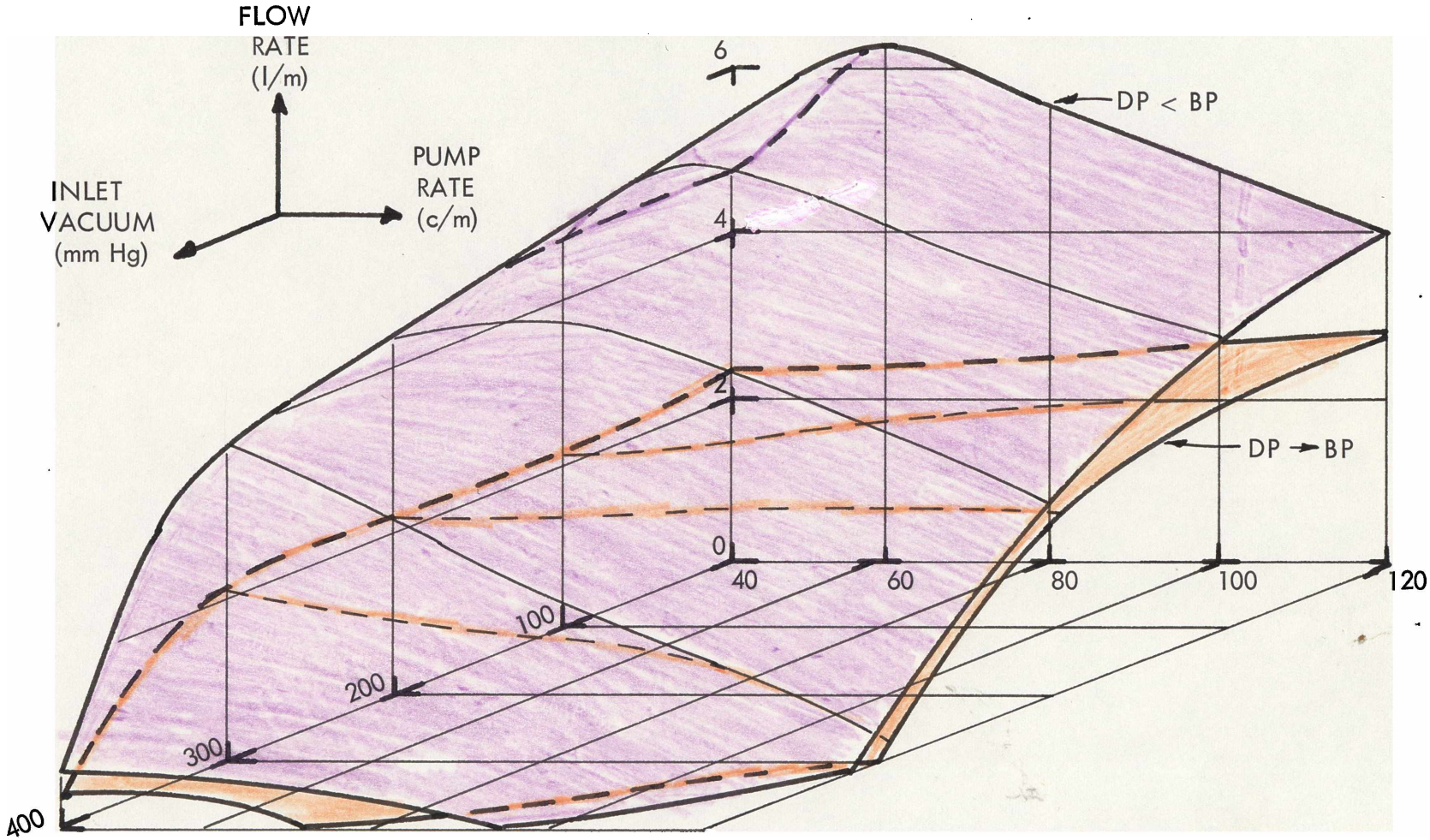


FIGURE 11

FLOW RATE, PUMP RATE, INLET VACUUM SURFACES, BP = 300 mm Hg

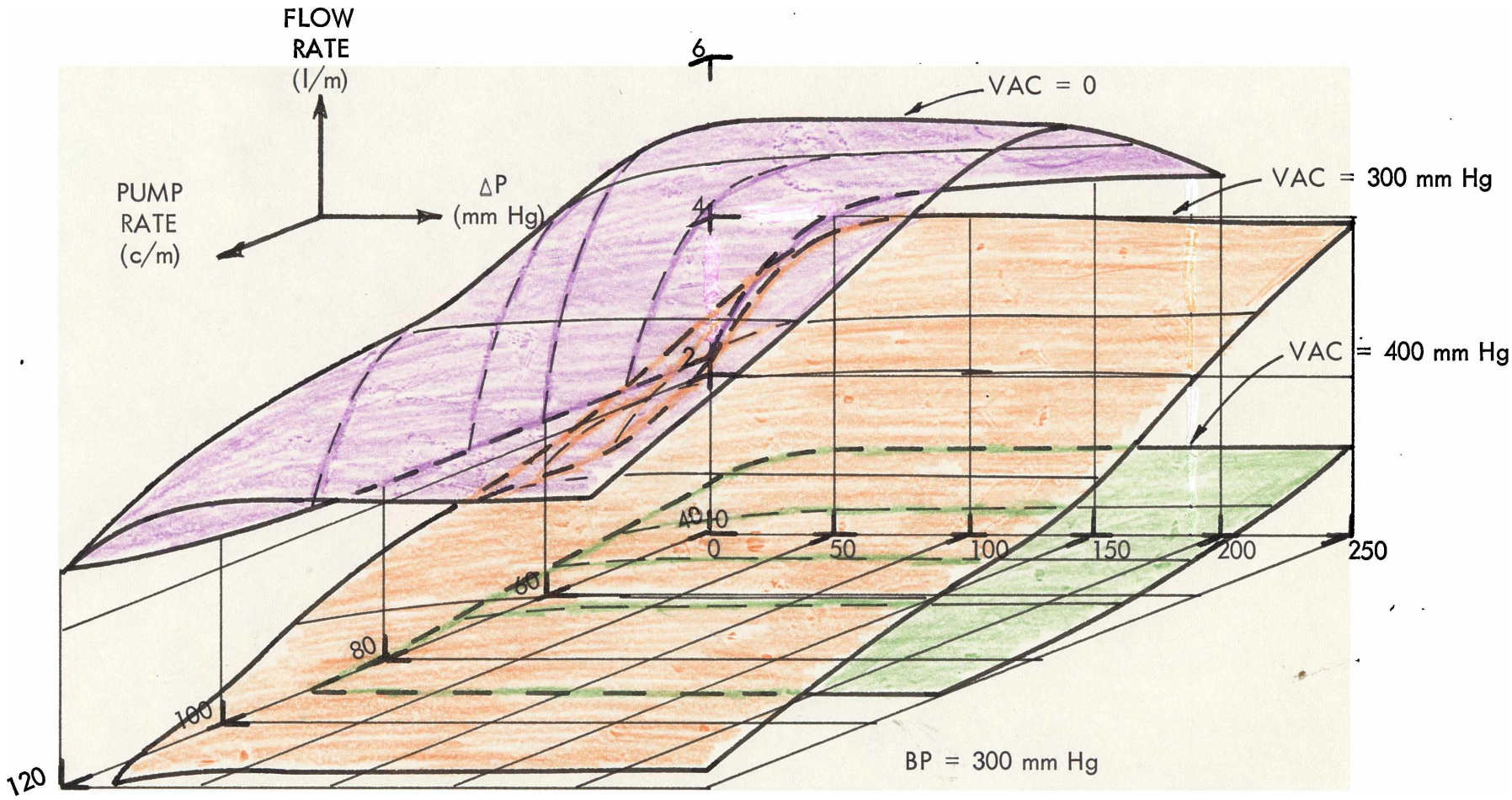


FIGURE 12

FLOW RATE, PRESSURE DIFFERENCE (BP-DP), PUMP RATE SURFACES

difficult to obtain. To help clarify the three-dimensional plots, planes can be passed through the surfaces parallel to the variables of interest and the information portrayed in two-dimensional form. Information from Figures 9, 10, 11, and 12 is thus shown in Figures 13, 14, 15, and 16.

Conclusions: From these data, the following conclusions about the pump can be made:

1. Flow rates in excess of 6 liters per minute were obtainable.
2. Flow rates were relatively independent of discharge pressure as long as the bag collapse pressure was at least 50 mm Hg higher than the discharge pressure.
3. Flow rates were not overly affected by increases in inlet vacuum until the inlet vacuum became greater than 300 mm Hg below atmospheric pressure. However, at vacuums above 300 mm Hg, flow rates were severely curtailed.
4. At lower vacuums, the curves showed peaks in the flow versus rate plane characteristic of many pumps. This peak disappeared with increasing vacuum.
5. Generally, flow rates decreased with increasing pump rates. When the bag and discharge pressures approached each other, though, flow rates at times increased with increasing pump rates.

In Vitro Hemolysis Studies

In order to compare blood trauma of this pump with other pumps, a series

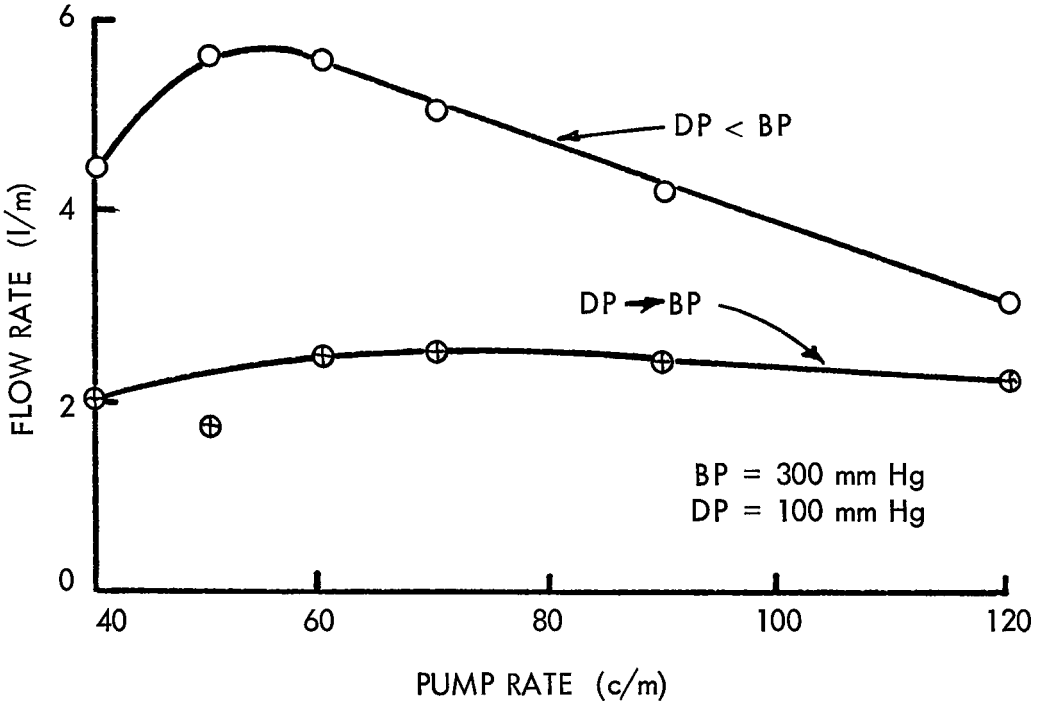


FIGURE 13

FLOW RATE VERSUS PUMP RATE

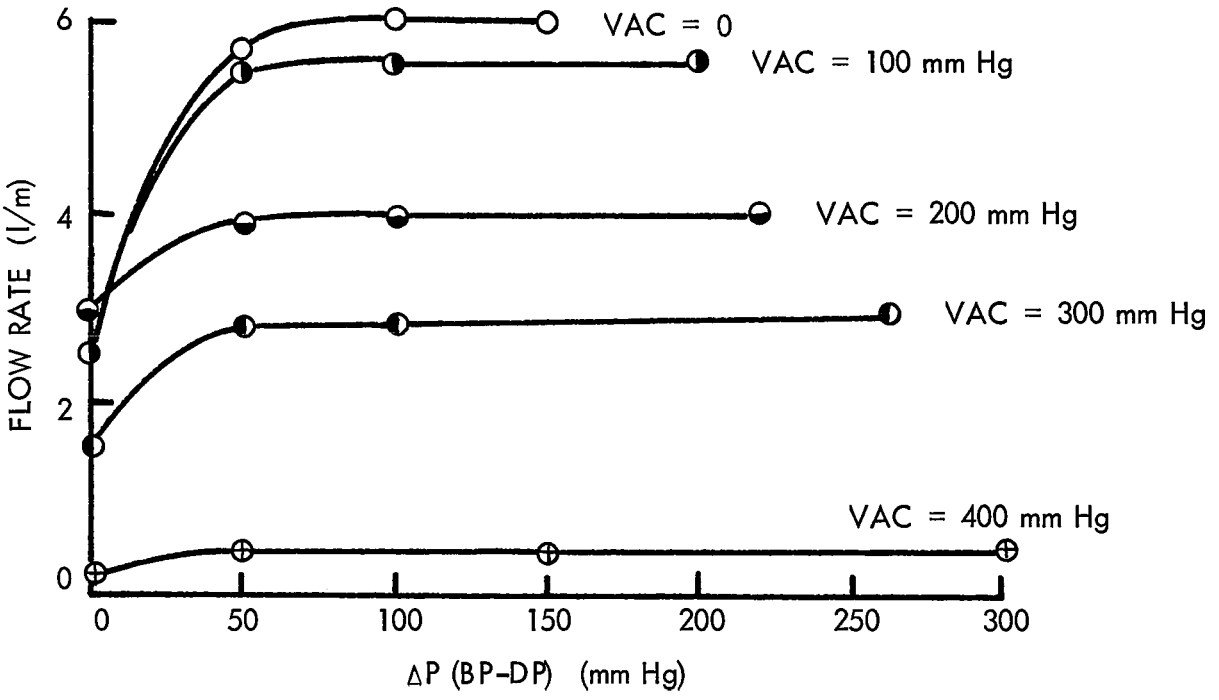


FIGURE 14

FLOW RATE VERSUS PRESSURE DIFFERENCE

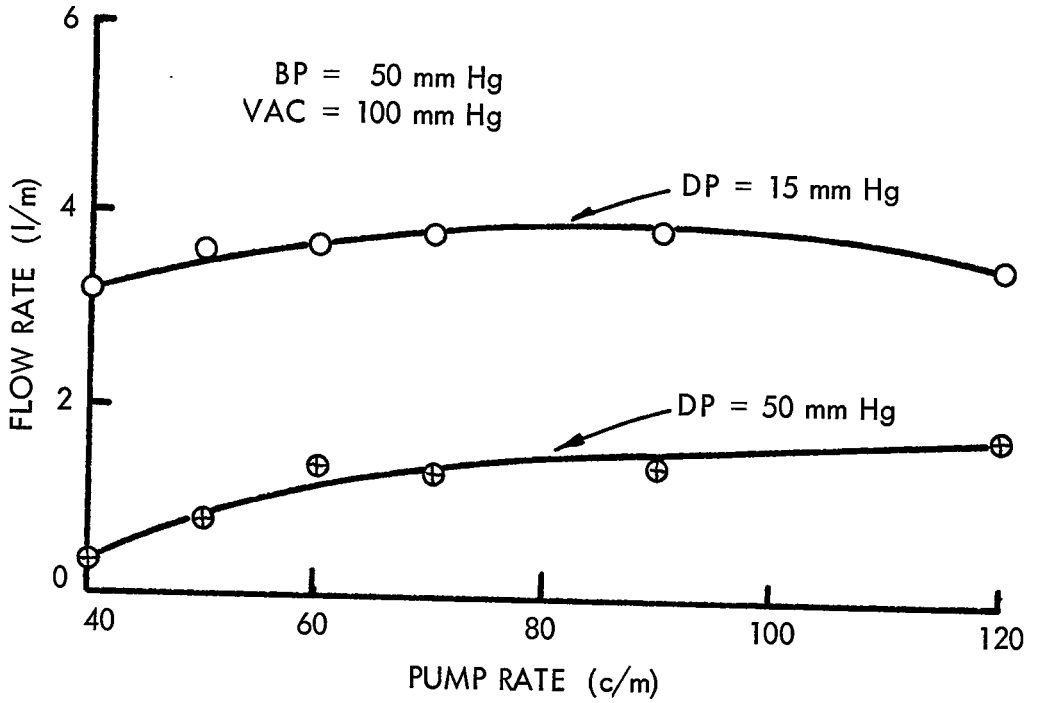


FIGURE 15

FLOW RATE VERSUS PUMP RATE

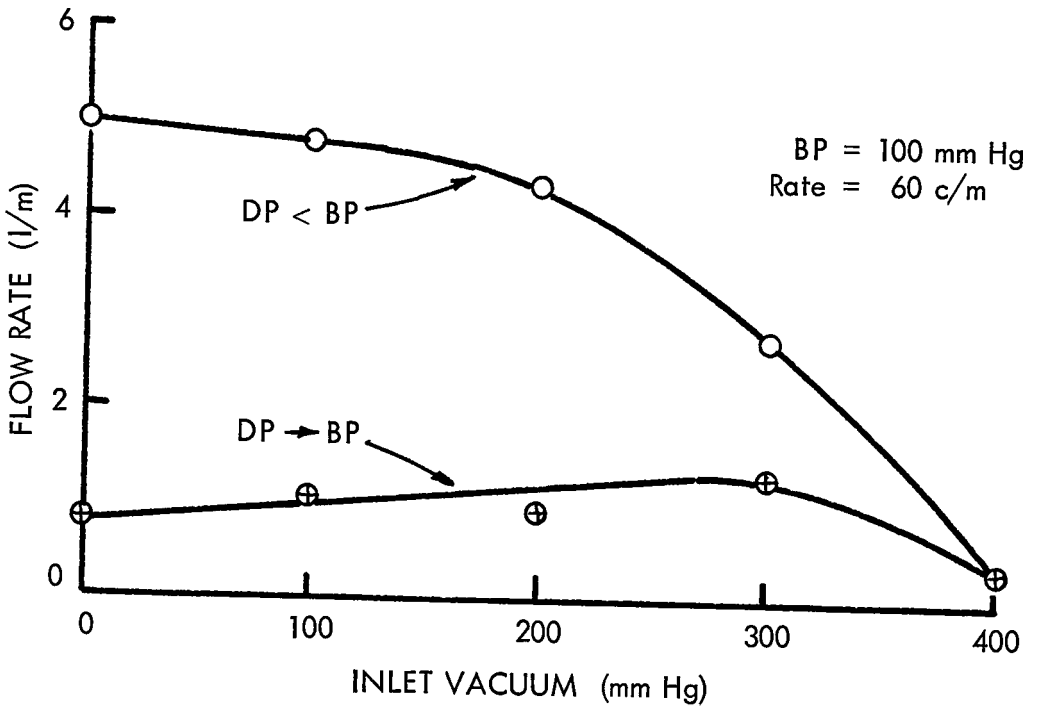


FIGURE 16

FLOW RATE VERSUS INLET VACUUM

of in vitro hemolysis studies was undertaken. The purpose of these tests was twofold: first, to obtain hemolysis rates of this pump in comparison with roller pumps; and, secondly, to determine if the parameters of the Geddes-Schuhmann method of left ventricular bypass are hemolytic, in particular the inlet restriction imposed by the small diameter ventricular catheter. The experimental apparatus is outlined in Figure 17. Basically, it consisted of a pump, blood reservoir, constant temperature bath and screw clamps for control of discharge pressure and inlet vacuum. One series included the use of the inlet catheter, and one series was run without this catheter. As a comparison, tests were run with both the previously described collapsible bag pump and with a roller-type pump (Sarns). Blood temperatures were maintained at 36°C.

Free plasma hemoglobin and blood hematocrit levels were checked at half-hour intervals during the test runs which lasted from 3 1/2 to 8 hours. Plasma hemoglobin levels were determined as follows: One ml of blood was drawn from the circuit, placed in a calibrated Wintrobe tube and spun in a centrifuge at 2400 rpm for 20 minutes. Following this, the hematocrit value was recorded and a plasma sample of either 20, 100 or 200 microliters, depending on the degree of hemolysis, was placed in 5 ml of Drabkin's Solution and compared by colorimetry (Bausch & Lomb Spectronic 20) at 540 millimicrons with a standard calibration solution. All blood used was freshly drawn, heparinized (30 mg/liter) canine blood which was diluted with 5% dextrose in 0.9% saline in the approximate ratio as that used in the in vivo experiments.

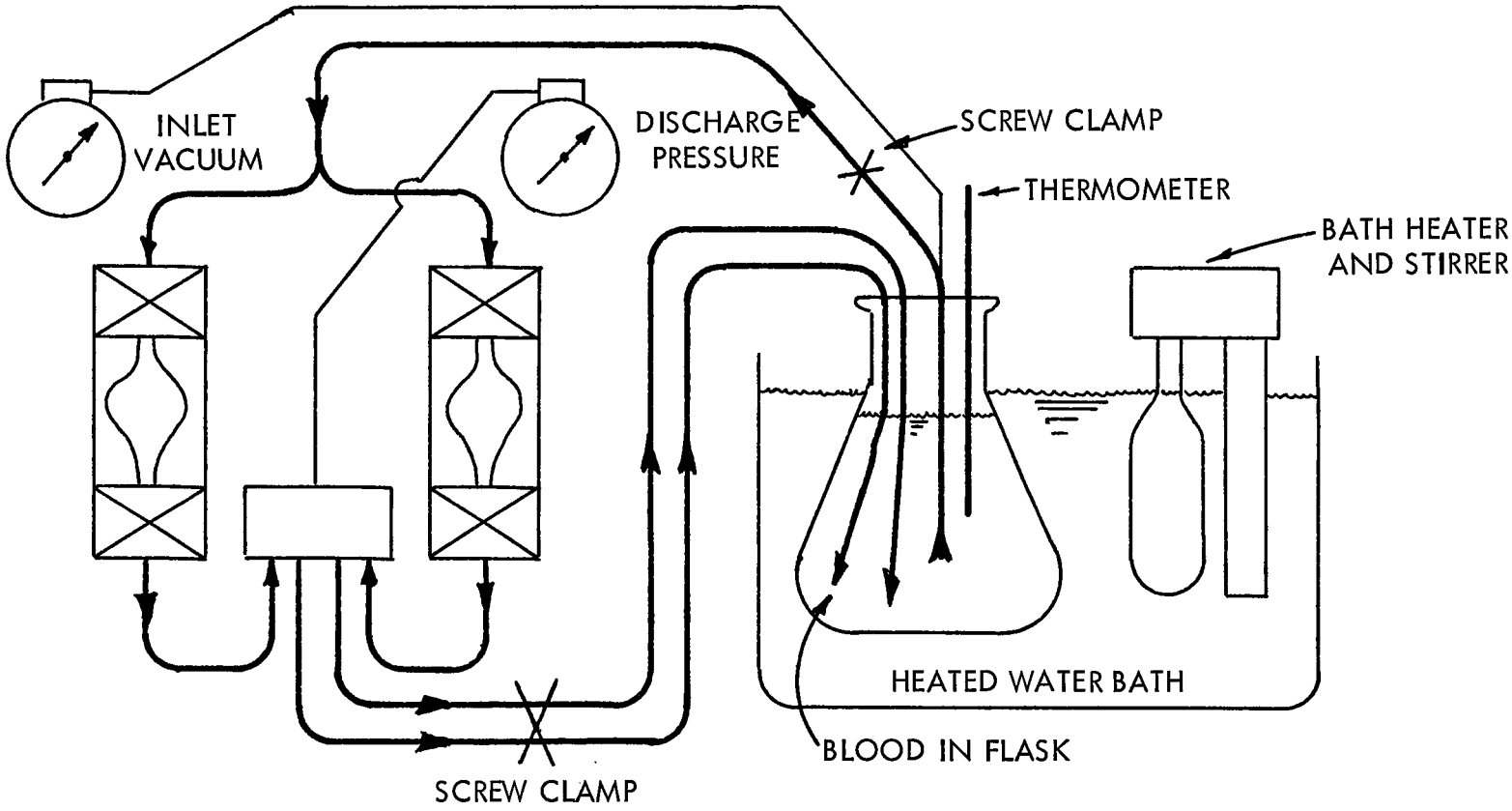


FIGURE 17

PUMP HEMOLYSIS STUDIES TEST APPARATUS

The results of these tests are shown in tabular form in Table II and graphically as plasma hemoglobin versus time in Figures 18 and 19. Whereas the number of tests was small and the hemolysis rates widely varied, the following conclusions can be drawn:

1. Hemolysis rates with the collapsible-bag type pump were definitely higher than with the roller pump. In face, they averaged 10 times the roller pump rates.
2. High flow rates did not appear to increase hemolysis rates.
3. Average hemolysis rates with a restricted inlet on the bag-type pump were significantly higher than rates without the restricted inlet, whereas with the roller pump no significant differences in hemolysis rates were noted, as shown in Figure 20. This indicates that the restriction itself was probably not hemolytic; rather, something associated with the dynamics of the bag pump, produced by the restricted inlet, was hemolytic. The problem was, in this investigator's opinion, as follows: With a restricted inlet, the bags in the pump operated in a nearly collapsed condition at all times. That is, they were occlusive. As has already been mentioned, occlusiveness is one of the primary causes of hemolysis. It then seems plausible that the occlusiveness of the bags, as caused by the restricted inlet, was the primary cause of hemolysis. Further evidence of this is shown in the fact that hemolysis rates were

TABLE II

IN VITRO HEMOLYSIS STUDIES I

Test No.	Pump Type	Inlet Vacuum (mm Hg)	Bag Condition	Test Duration (hours)	Flow Rate (l/m)	mgms Hb Liberated per hour	Index of Hemolysis	Blood Hct.
H 1	Old Bag	0	Part. Collapse	3 1/2	6	96	0.755	23
H 2	Old Bag	0	Open	6 1/2	3/4	52	1.14	22
H 3	Old Bag	400	Collapsed	7 1/2	1.5	247	2.04	19
H 4	Old Bag	400	Collapsed	7	1.1	121	1.24	28
H 5	Roller	0	---	6 1/2	2	13.1	0.098	19
H 6	Roller	350	---	6 1/2	1.2	20 (0-2½ hrs) 3.3 (2½-6½ hrs)	0.244 0.041	20 20
H 7	Roller	190	---	7	0.7	6.0	0.095	27
H 8	Roller	0	---	7	3	3.5 (0-6 hrs) 50 (6-7 hrs)	0.0188	26
H 9	Old Bag	310	Collapsed	6	1.5	164	1.21	30
H 10	Roller	340	---	6	1.2	12.5	0.133	30

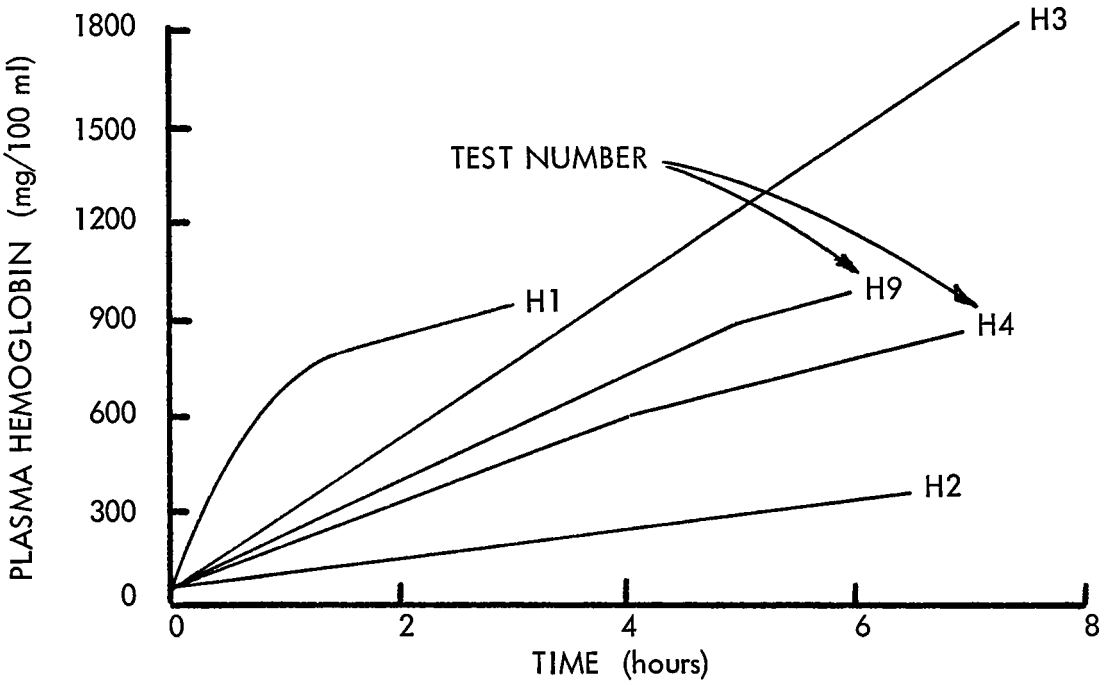


FIGURE 18

PLASMA HEMOGLOBIN VERSUS TIME, BAG PUMP

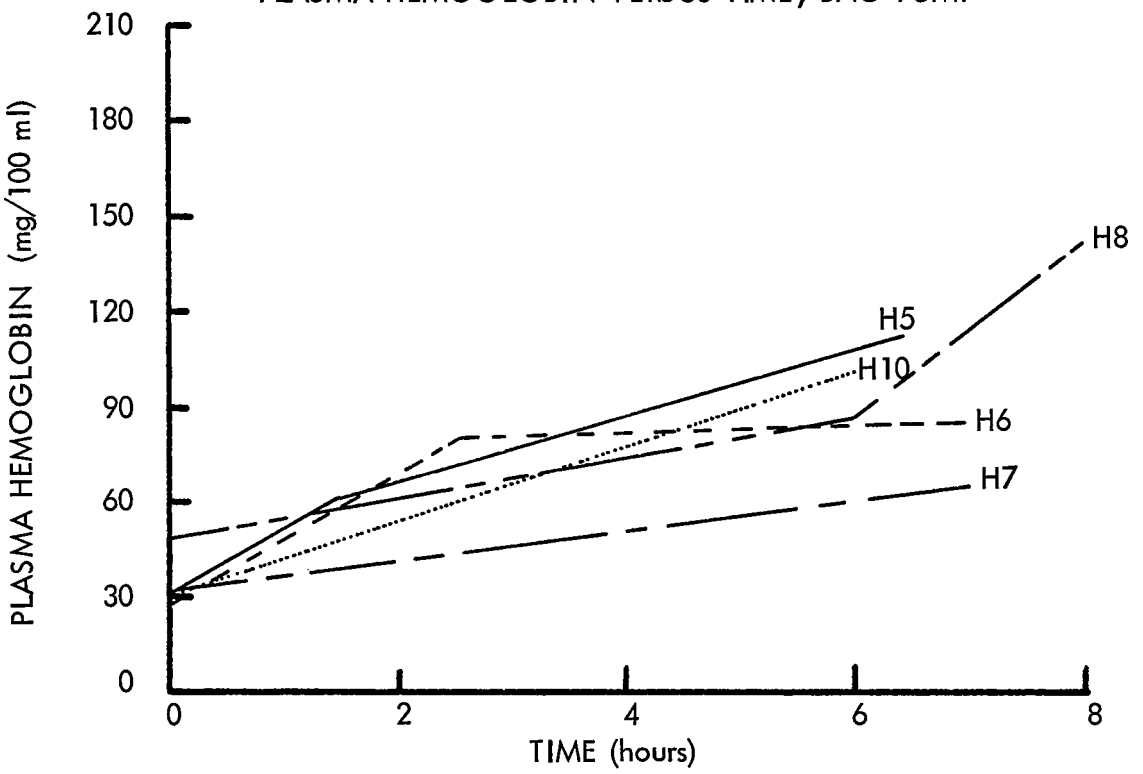


FIGURE 19

PLASMA HEMOGLOBIN VERSUS TIME, ROLLER PUMP

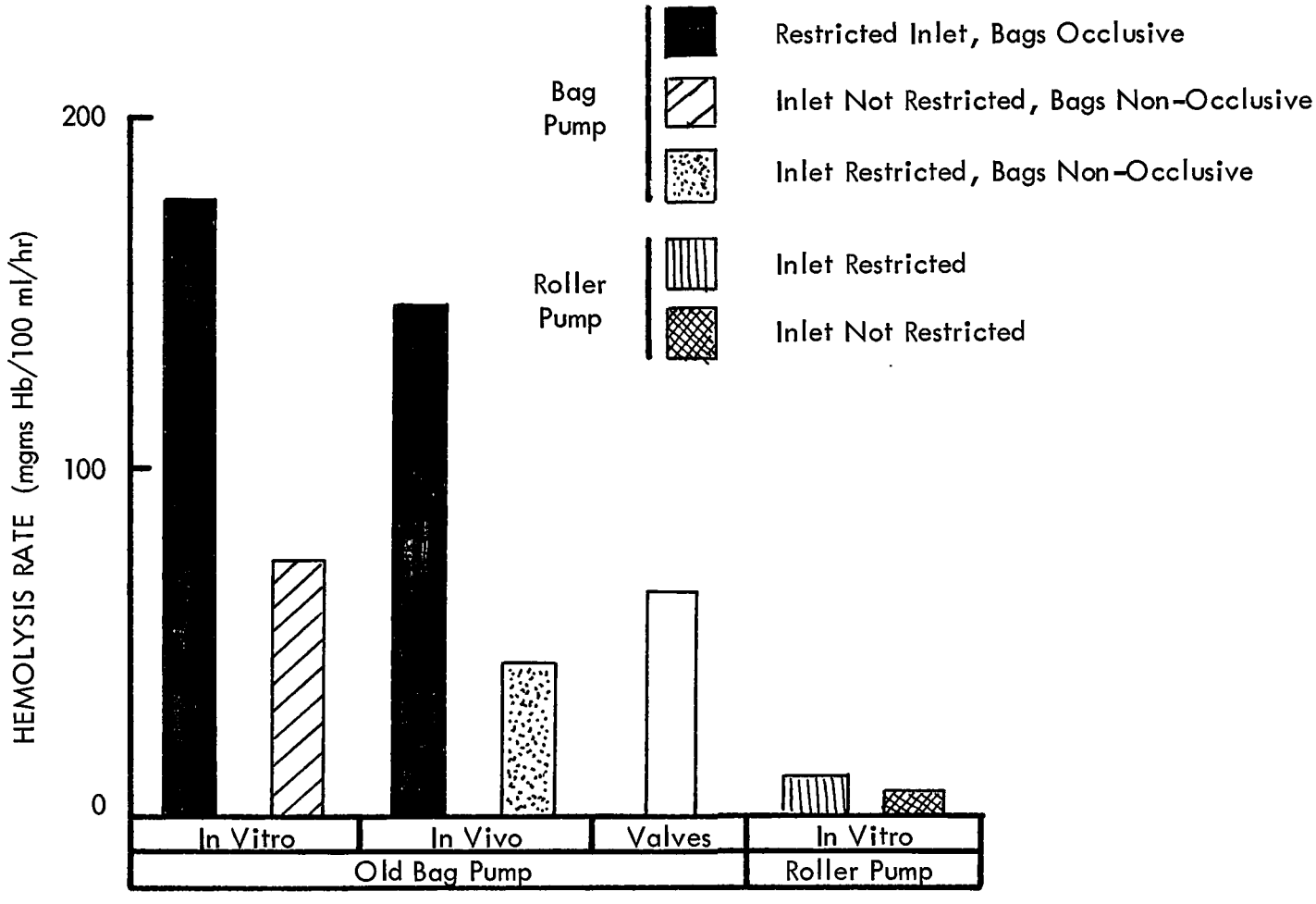


FIGURE 20

HEMOLYSIS STUDIES SUMMARY

decreased when the bags were operated in a non-collapsed condition. The fact that hemolysis continued, even with the bags operated non-occlusively, indicates that other factors were present. The indication now is that the active valves were hemolytic. Since their operation requires occlusiveness, this is not surprising.

To further clarify this point, a short series of hemolysis studies involving only the valves was undertaken. In these studies, freshly drawn, heparinized canine blood was placed in the pumping chambers and the valves actuated. The pump bag operated as a flexible reservoir and permitted withdrawal of blood samples. The total system was immersed in a temperature-controlled water bath and the same experimental procedure followed as in the other hemolysis studies. The results of these tests are shown in Table III and summarized in Figures 20 and 21. These studies indicate that, with the pump non-occlusive, most of the hemolysis was due to the valves. In addition, Figure 20 shows that the hemolysis rate was a function of the valve collapse pressure and increased with increasing pressure.

The question might now be asked, if the roller pump is occlusive, why aren't its hemolysis rates higher? There are two reasons: first, the occlusive area is small in relation to the bag pump; and secondly, the occlusiveness occurs in a different manner.

TABLE III

VALVE HEMOLYSIS STUDIES

Test No.	Test Duration (hours)	Valve Rate (c/m)	Valve Pressure (mm Hg)	mgms Hb Liberated per hour	Hb Liberation Rate (corrected)	Blood Hct
VH 1	5	66	700	402	76.6	25
	5	66	700	402	76.6	25
VH 2	2	66	100	32.5	6.5	27
	2	66	200	80	17.5	27
	2	66	300	75	15	27
	2	66	400	112.5	24.5	26
	2	66	500	97.5	19.5	26
	2	66	500	95	20.7	26
	2	66	600	142.5	31	26
	2	66	700	195	39	26

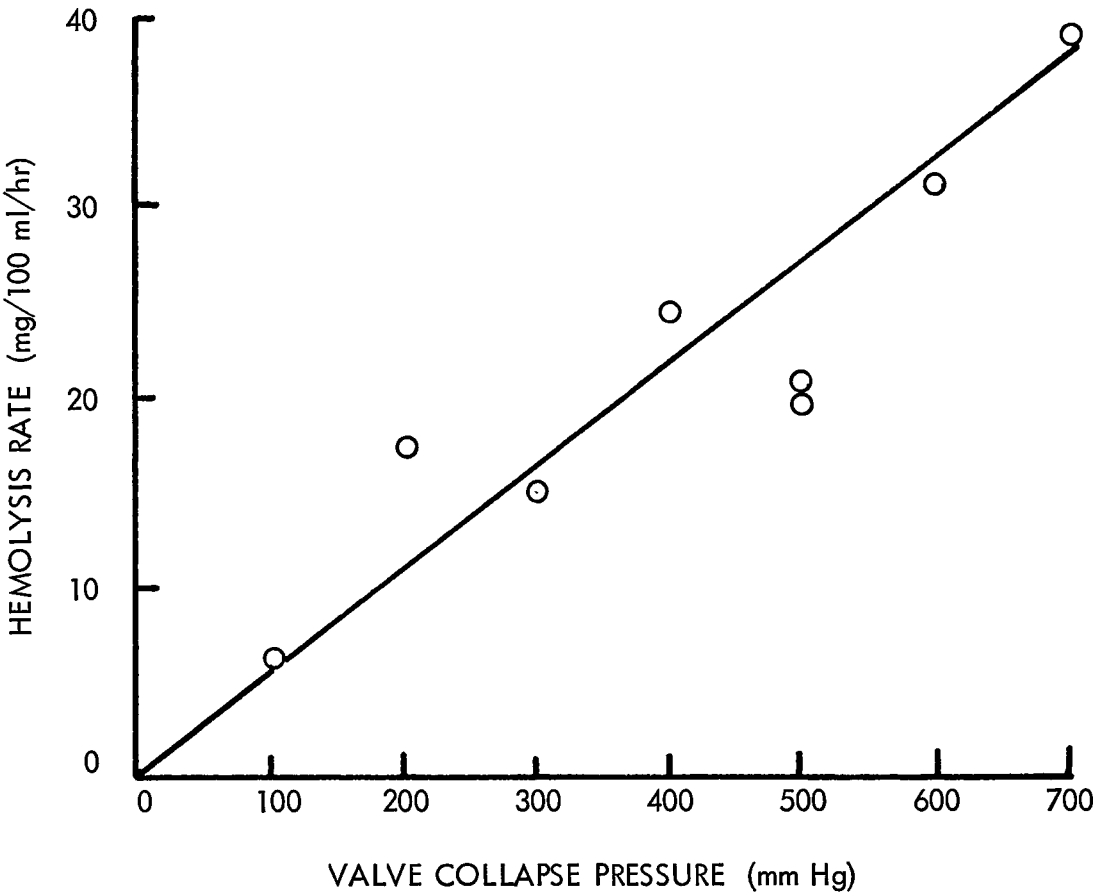


FIGURE 21

HEMOLYSIS RATE VERSUS VALVE COLLAPSE PRESSURE

In the roller pump the roller moves around so as to push the red blood cells out in front and away from the area of contact. The opposite is true in the bag pump as the bag collapses in a random manner trapping and squeezing the red blood cells.

5. Variations from animal to animal in red blood cell fragility are evident as indicated by the wide variation in hemolysis rates for similar pumping conditions. This complicated the obtaining of quantitative information.

In Vivo Studies

In conjunction with the evaluation experiments of the Geddes-Schuhmann method, a series of studies was performed in which the previously described pump was used with living animals. This method is described in Appendix A. The dogs weighed from 18 to 25 kg and were anesthetized with sodium pentobarbital (30 mg/kg). Clotting was inhibited with heparin (2.5 mg/kg). Most of these bypass procedures were carried out with fibrillating ventricles to insure total circulatory maintenance. Parameters of interest in this study were plasma hemoglobin level, blood hematocrit and total blood flow rate (cardiac or pump output). Other variables were measured but are not presented here as the subject of this investigation is pump performance and not necessarily physiological response to pumping. The variables studied included blood pH, blood gas measurements (pO_2, pCO_2), blood bicarbonate levels and others.

Pumping durations in these experiments ranged from one to more than four hours. Plasma hemoglobin levels and blood hematocrit were determined prior to pumping, at half-hourly intervals during pumping, and at hourly intervals following pumping. The method of measurement was the same as described under the in vitro experiment section. Cardiac and pump output values, as determined by the saline conductivity method [91, 92], were taken at varying intervals during the experiment. A summary of the results of these tests is shown in Table IV.

The following observations can be made from these experiments:

1. Maximum flow rate achieved in the in vivo experiments was 1.7 liters per minute.
2. Plasma hemoglobin levels rose more rapidly than desirable. Rates of plasma hemoglobin collection varied from 30 to 220 mg per hour per 100 ml of plasma.
3. Again, as was the case with the in vitro tests, hemolysis rates with the bags operating in a collapsed manner were higher than with the bags non-occlusive. Since hemolysis rates increase with the bags in a collapsed state, it would be desirable to operate with the bags nearly full. In practice, this is difficult to achieve, and it requires reducing the bag collapse pressures to near-discharge pressure with the resultant reduction in flow.
4. At times, for unexplained reasons, the inlet vacuum increased to

TABLE IV

IN VIVO PUMP STUDIES I

Test No.	Bag Condition	Pumping Duration (hours)	Flow Rate (l/m)	Blood Hct.	mgms Hb Liberated per hour	mgms Hb Cleared per hour	Remarks
64-P1	Collapsed	1/2	Low	--	---	--	Insufficient Flow
		1/4	Low	--	---	--	
65-P2	---	2	Good	49	---	--	Death - Spontaneous Fibrillation
66-P3	---	1	Good	30	---	--	Good Recovery
70-P4	Collapsed	1 1/6	0.8	45	220	48	Good Recovery
71-P5	Collapsed	3	1.4	30	166	67	Poor Recovery - Congestive Failure
72-P6	Collapsed	2 1/4	Low	48	105	None	Death - Unable to Defibrillate
73-P7	---	---	---	35	---	--	Unable to Pump - Heart Worms
74-P8	Open	2 1/3	1.2	42	30	3.2	Good Recovery
75-P9	---	1/2	Low	37	---	--	Poor Oxygenation

TABLE IV (Continued)

IN VIVO PUMP STUDIES I

Test No.	Bag Condition	Pumping Duration (hours)	Flow Rate (l/m)	Blood Hct.	mgms Hb Liberated per hour	mgms Hb Cleared per hour	Remarks
76-P10	---	1/4	Nil	36	---	--	Death - Congestive Failure (2 hrs)
		1 1/6	Fair		---	--	
77-P11	---	---	---	--	---	--	Death - Technical Error
78-P12	Collapsed	1	1.7-0.4	41	103	None	Good Recovery
79-P13	Part. Collapse	2 1/2	0.4	42	60	None	Death - Shock-Like Condition (4 hrs)
80-P14	Collapsed	3 3/4	1.0	38	142	72	Death - Unexplained (5 hrs)
81-P15	Open	1 1/3	1.2	34	39	1.7	Good Recovery (non-fibrillatory expt.)

400 mm Hg and above. As shown from the pump curves, this caused a severe decrease in pump flow rates. The causative factor was difficult to pinpoint, but the most likely cause was the occlusion of all or part of the catheter tip in the ventricle. This could have been due to the endocardium, cordae tendinae, valve leaflets or other material, such as thrombus formation at the tip. The solution to this problem was not found and is still not clear.

Discussion

On the basis of these observations, the time had come to make some decisions concerning the design and operation of the collapsible-bag pump and to attempt its redesign to improve on its properties. The following summarizes characteristics of the pump tested:

1. Flow rates sufficient to maintain total circulation in mongrel dogs for times exceeding four hours were obtainable.
2. Hemolysis rates were higher than desirable, due to the occlusive nature of the pumping bag and active valves.
3. The pulsatile nature of this pump was not as great as desired. In fact, the pulse measured in the animal's femoral artery was almost indistinguishable, save a small transient fluctuation at the onset of pump discharge. This is shown in the sample Physiograph record of Figure 22.

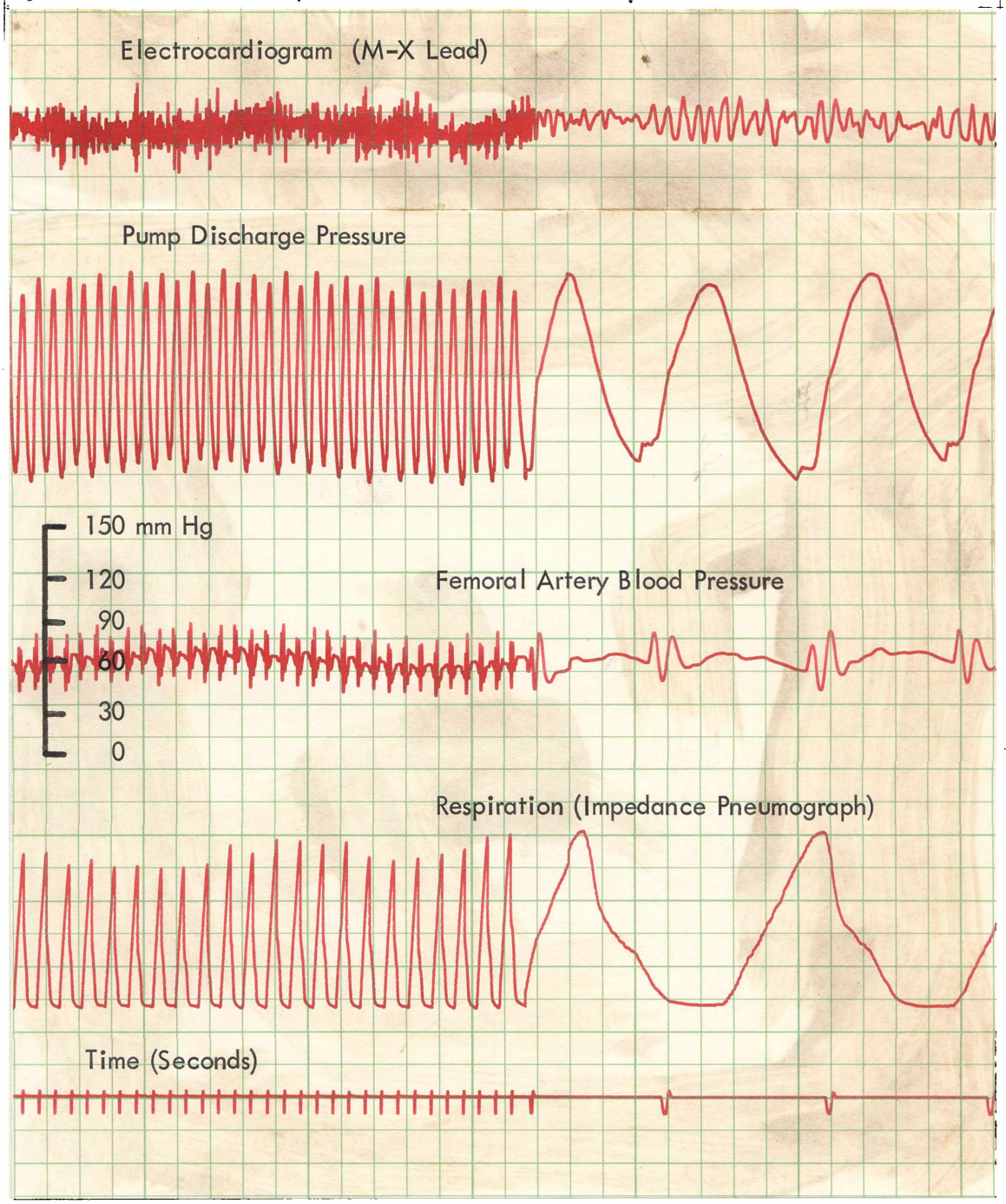


FIGURE 22

SAMPLE PUMPING PHYSIOGRAPH RECORD, EXPT. 66-P3

4. The output of the pump was poor above 60 cycles per minute. When it becomes desirable to synchronize the pump with the beating heart, as with assisted circulation, much higher pump rates will be necessary. An animal under sodium pentobarbital anesthesia normally has a heart rate near 180 beats per minute and somewhat higher. It is, of course, possible to infuse on alternate cycles or even every third or fourth cycle although this is not as desirable as every cycle.

A NEW PUMP

Pump Design

The first step in the redesign of the pump was to decide what improvements were necessary and what could be done to implement them. Most of the problems encountered with the first pump model are given in the previous section and represent the areas for improvement. These problems were: (1) reduced flow at pump rates higher than 70 strokes per minute, (2) low flow rates at high inlet vacuums, and (3) high hemolysis rates.

(1) Low Flow at High Pump Rates

This was probably the easiest problem to approach for several reasons. First of all, there was not too much that could be done without totally altering the pumping concept. Also, this problem was not as great as the others and as

a result did not require dramatic improvement. The two possible methods of increasing the pumping rate were associated with the pump itself and with the control system. By simply reducing the size of the pumping chamber, a resultant increase in pumping frequency should be seen because less volume of the control fluid (air) need be moved back and forth for each pump cycle.

Changes in the control system could include enlarging tubing size to reduce fluid friction, reducing tubing lengths, removal of other flow restrictions and increasing fluid pump capacities. As the initial pump capacities were larger than necessary, not much could be gained here. Room for some improvement in the flow restriction problem is possible however.

(2) Low Flow at High Inlet Vacuum

This problem was related to the fact that pumps with pulsatile outflows have pulsatile inflows. Maximum inflow, however, would occur if a steady, non-pulsatile inflow could be maintained. This was partially helped by the push-pull parallel pump configuration, but further improvement was desirable. In addition, the push-pull pump yielded minimal pulsations in flow.

(3) High Hemolysis Rates

It has been found that the major cause of hemolysis in the first pump model was occlusion of the bags (at higher inlet vacuums) and the active valves. This, then, was the first place to start to reduce hemolysis rates. Since active valves are by nature occlusive, all that could be done was to reduce the

occlusive area. If the same internal diameter was to be retained, the only occlusive area reduction method was to shorten the valve length.

As far as the pump bag is concerned, the best method was to eliminate the occlusive pumping nature. This means that the bag should not be allowed to totally collapse. Furthermore, this requires a variable duty cycle which is not feasible with the push-pull parallel pump in which each bag is operated alternately and hence requires a 50% duty cycle control system. Another improvement, if it was not possible to eliminate totally the occlusiveness, would have been to reduce the occlusive surface area.

Thrombus Formation

Thrombus formation in the original pump was not serious with heparinization and, thus, did not require major modification. However, it should be kept in mind that stagnation areas should be eliminated, surfaces should be kept smooth and material interfaces be kept few in number.

A New Pump

How to combine all of the above concepts into a single pump was the next problem to be considered. To gain the variable duty cycle capability required a single-bag pump. This type of pump would also yield a more pulsatile flow. Retaining the straight-through, streamlined pump-chamber design should keep thrombus formation low. If a cylindrical pump chamber were used, instead of the original elliptical shape, fabrication would be much easier and

fewer material seams would be present. The desire was to maintain the volume-surface area ratio as high as possible. For a cylinder in which only the cylindrical portion is occlusive, this ratio is

$$\frac{\text{Volume}}{\text{Surface}} = \frac{\pi r^2 h}{2\pi rh} = \frac{r}{2} \quad \text{(cylinder)} \qquad \text{(IV-1)}$$

where r = cylinder radius,
h = cylinder height.

Thus, as large a radius as practical should be used. This ratio is higher than that of a sphere which has the ratio

$$\frac{\text{Volume}}{\text{Surface}} = \frac{4/3\pi r^3}{4\pi r^2} = \frac{r}{3} \quad \text{(sphere)} \qquad \text{(IV-2)}$$

where r = sphere radius.

A reduction in pump size to approximately one-half the original should permit a higher pump frequency and still maintain adequate stroke volume.

Reducing valve-induced hemolysis by shortening the valves was not as easy as expected since excessive reduction in length impaired the competency of the valves. Several valve designs were tried, mostly with poor results. The final design was one of a simple cylinder of the Silastic material slightly pre-formed to the tricuspid shape in which is collapses. This style of valve was attached to the bag in the new pump instead of being a separate part. This placement further reduces seams and facilitates assembly of the pump chamber.

The design changes for all of these problems, except the high inlet vacuum flow, have now been accomplished. Here, an altogether new idea

was needed. What was necessary was a method of regulating the blood inflow so that blood would be drawn into a reservoir at all times. The pumping chamber could then be filled from the reservoir, as it is in the biological heart. This idea has been tried before by Wesolowski and Welch [97], but they experienced trouble in controlling the blood level in the reservoir. They achieved the steady withdrawal by merely applying suction directly to the space above the blood in the reservoir. Control of blood volume was a feedback sensor which regulated pump flow rate in relation to reservoir fluid level.

The vacuum reservoir idea should solve the problem and was conceived without prior knowledge of Wesolowski's work. In this application, rather than applying the vacuum directly to the blood, it was proposed that it be applied to the outside of a bag similar to the pump bag of the old style pump. Control of this vacuum would be manual by allowing air into the vacuum line. When correctly adjusted, the reservoir would partially empty during filling of the pump chamber and fill during pump ejection. Since there is also a vacuum on the pump bag during filling, a steady withdrawal of blood from the animal is the result. It is important that the vacuum on the pump bag be somewhat greater than the reservoir bag so that blood will be transferred from the reservoir to the pump. Physically, the reservoir is attached directly to the top of the pump and has a volume approximately twice that of the pump bag. A sketch of the pump and the control circuitry is given in Figures 23 and 24. Also, in Figure 25 a sketch of the final pumping chamber and valve design is given.

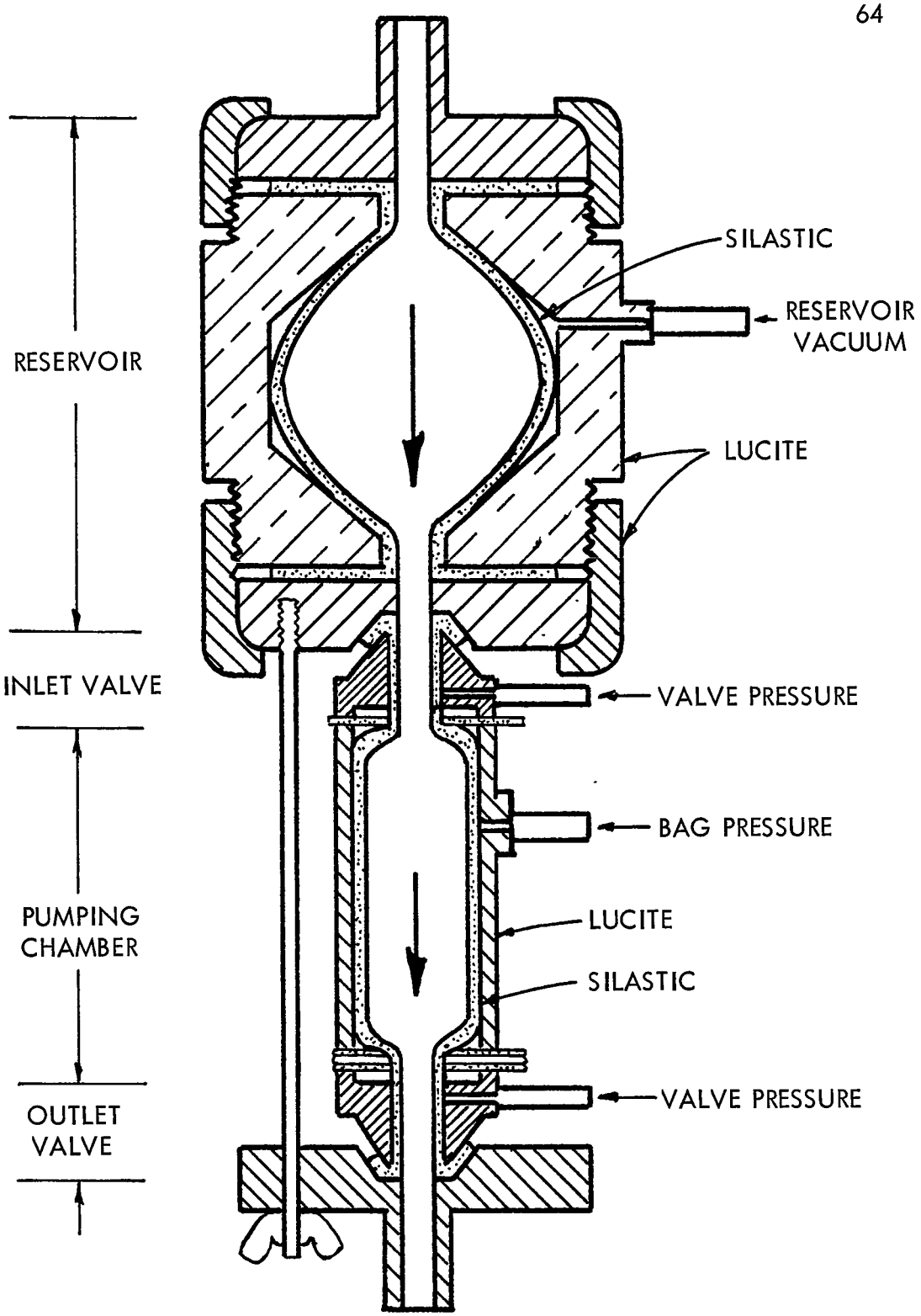


FIGURE 23

RESERVOIR PUMP CROSS SECTION

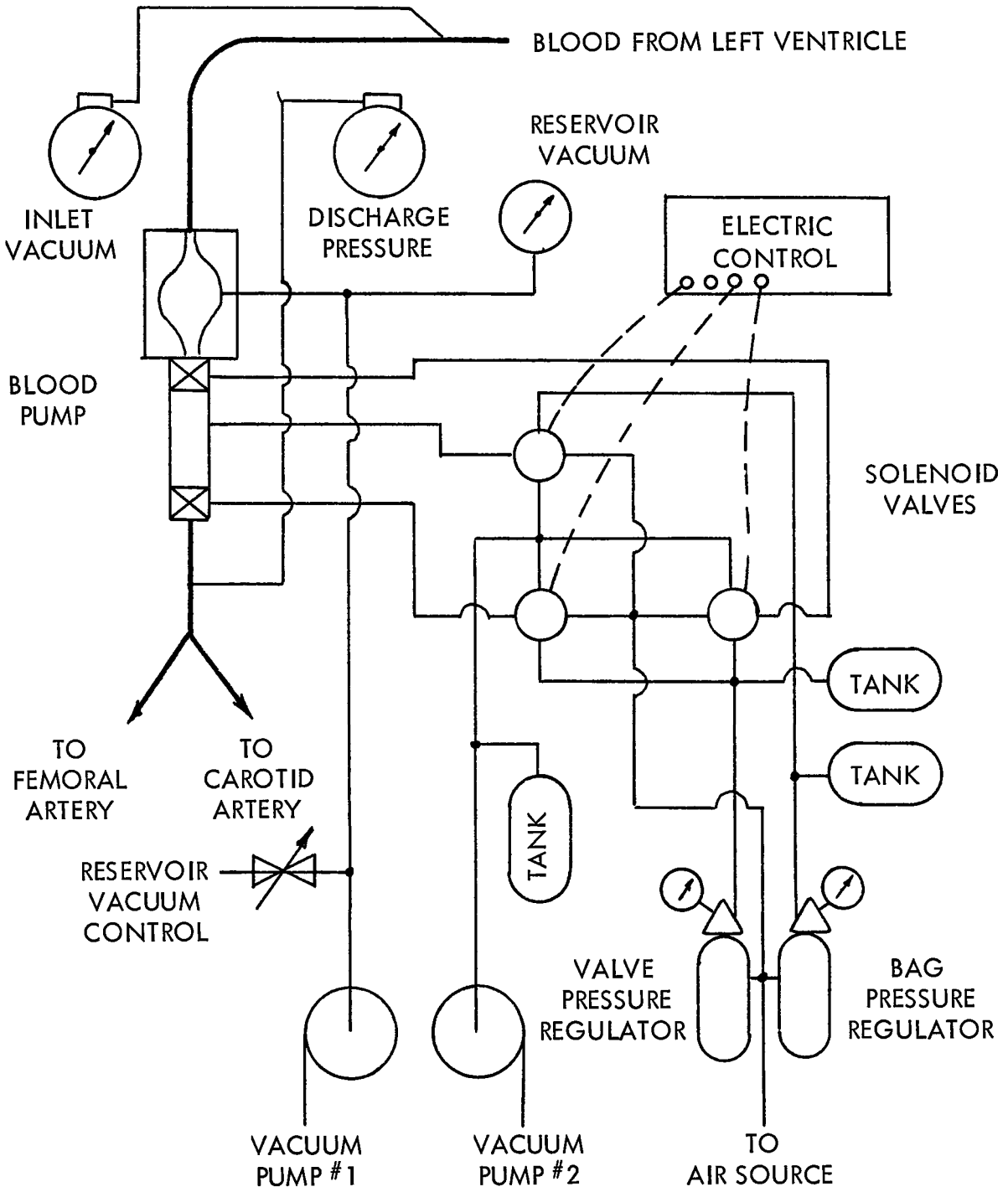


FIGURE 24

RESERVOIR PUMP BLOCK DIAGRAM

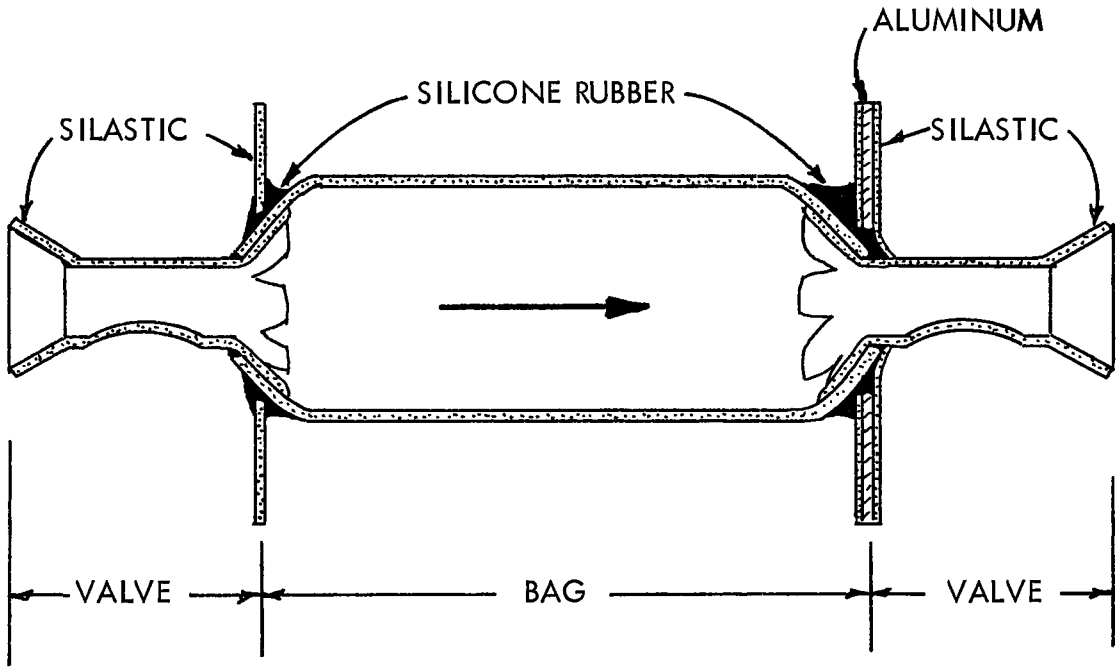


FIGURE 25

NEW PUMPING CHAMBER DESIGN

Evaluation of New Pump

Again, as was previously done, testing was carried out in three steps: in vitro flow tests using water, in vitro hemolysis studies and in vivo tests with animals.

In Vitro Flow Studies

Using the same experimental procedure and apparatus as in the previous flow studies, flow tests on the reservoir pump were run. The variables and values used for this pump are listed in Table V.

TABLE V

RESERVOIR PUMP FLOW STUDY PARAMETERS

Variable	Range Tested
Pump Rate (R)	40, 60, 80, 120, 160, 200 c/m
Bag Pressure (BP)	100, 200, 300 mm Hg
Inlet Valve (IV)	0, 100, 200, 300, 400 mm Hg Vac.
Discharge Pressure (DP)	0, 100, 200, 300* mm Hg
Duty Cycle	33%, 50%
Reservoir State	Active, Non-Active

* and 50 mm Hg less than BP.

The tests conducted with the parameters shown in Table V yielded an enormous amount of data, more than practical to present in tabular form. Instead, certain characteristic three-dimensional plots are shown in Figures 26, 27, 28, 29 and 30. In these figures, two axes are shown: Flow Rate vs. Pump Rate vs. Inlet Vacuum and Flow Rate vs. Pressure Difference vs. Pump Rate.

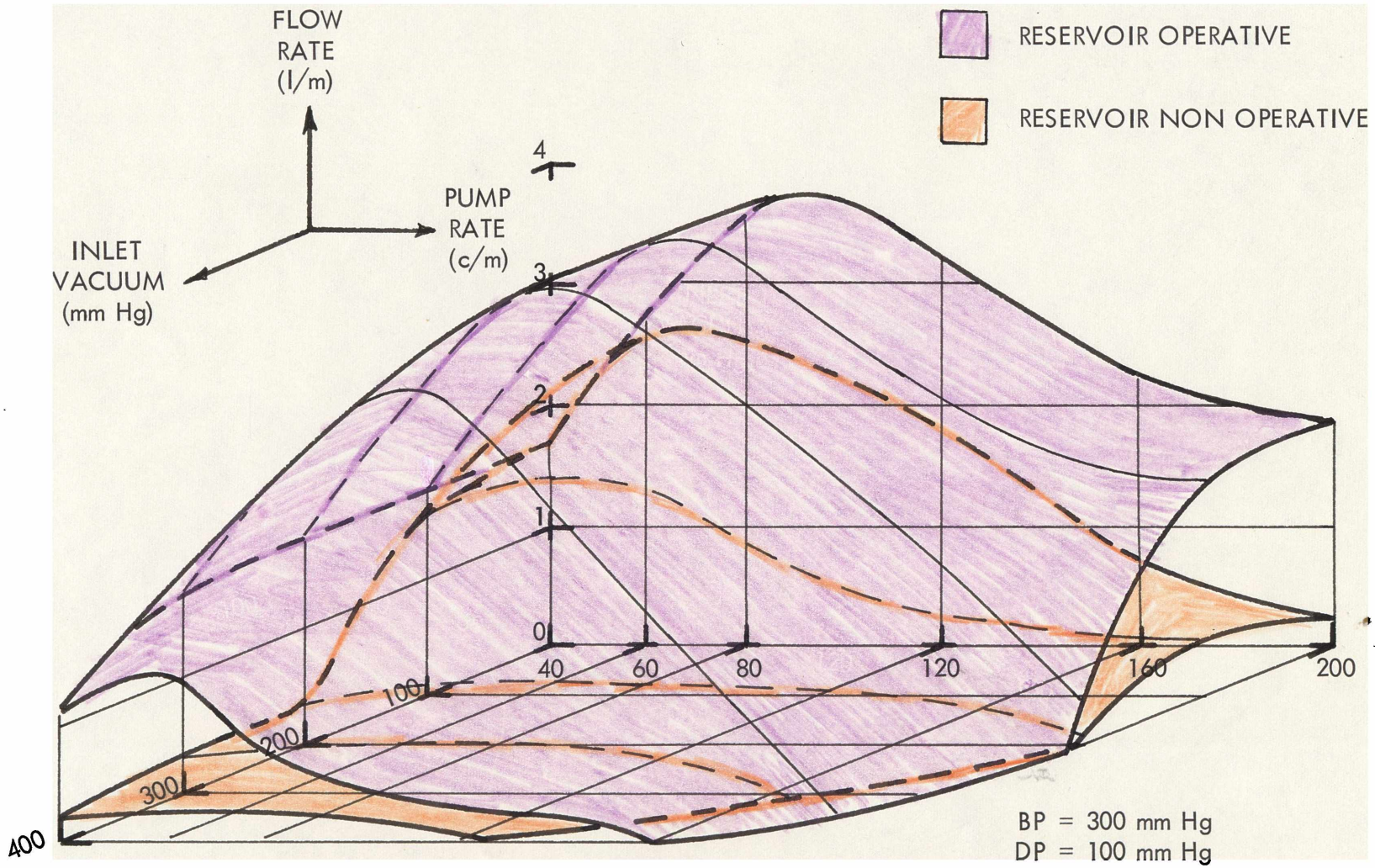


FIGURE 26

FLOW RATE, PUMP RATE, INLET VACUUM SURFACES, DUTY CYCLE 50%

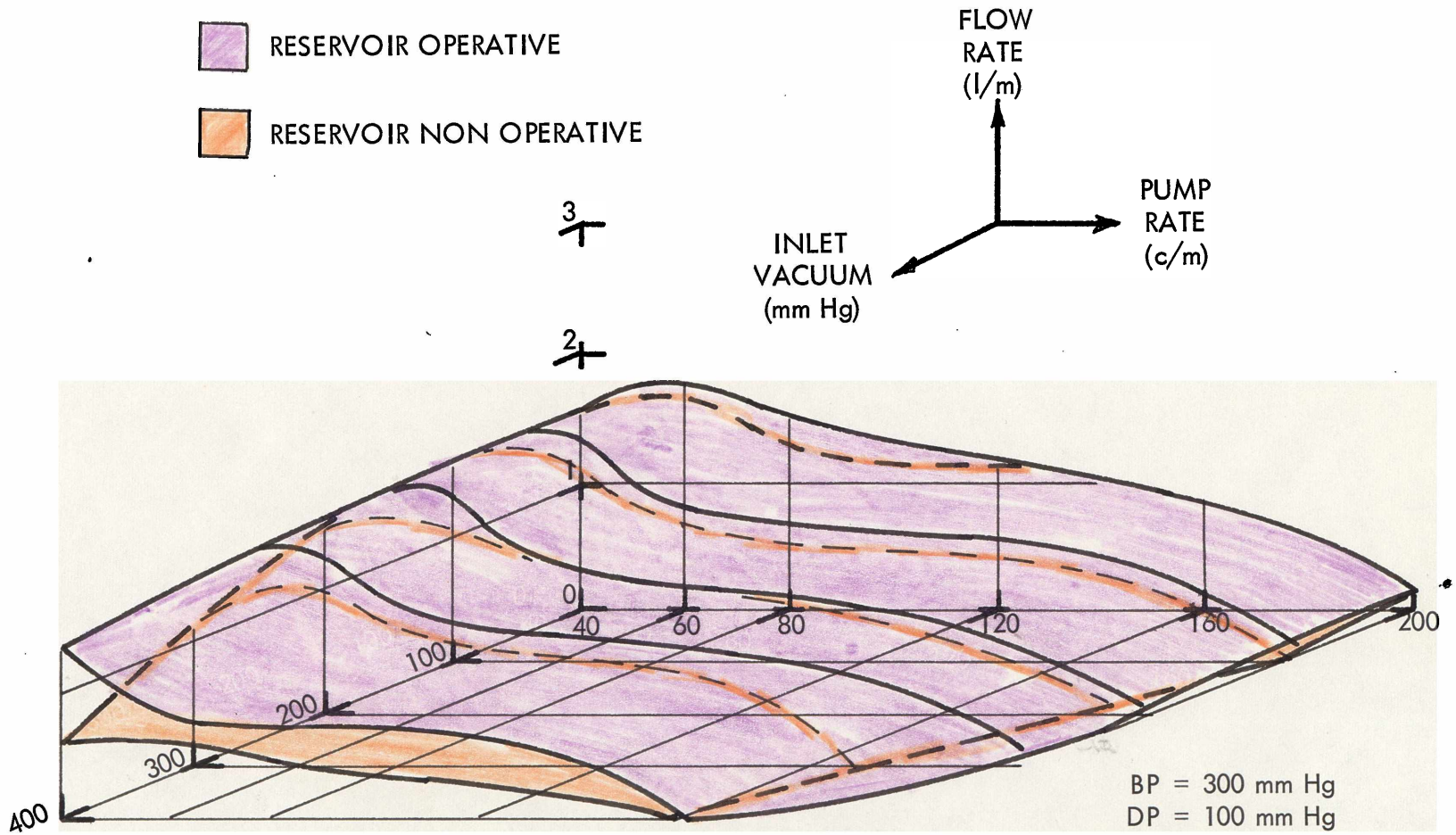


FIGURE 27

FLOW RATE, PUMP RATE, INLET VACUUM SURFACES, DUTY CYCLE 33%

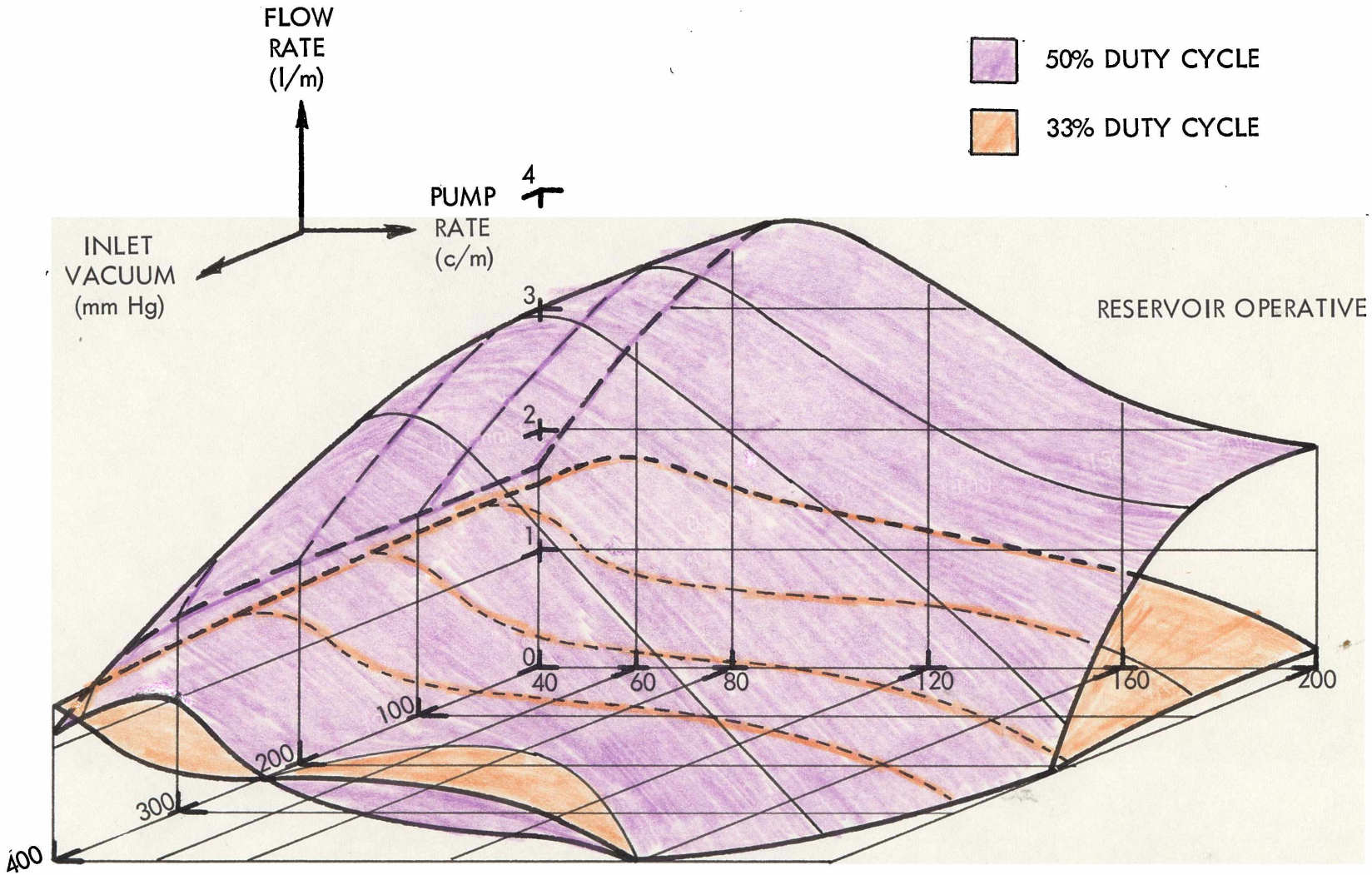


FIGURE 28

FLOW RATE, PUMP RATE, INLET VACUUM SURFACES

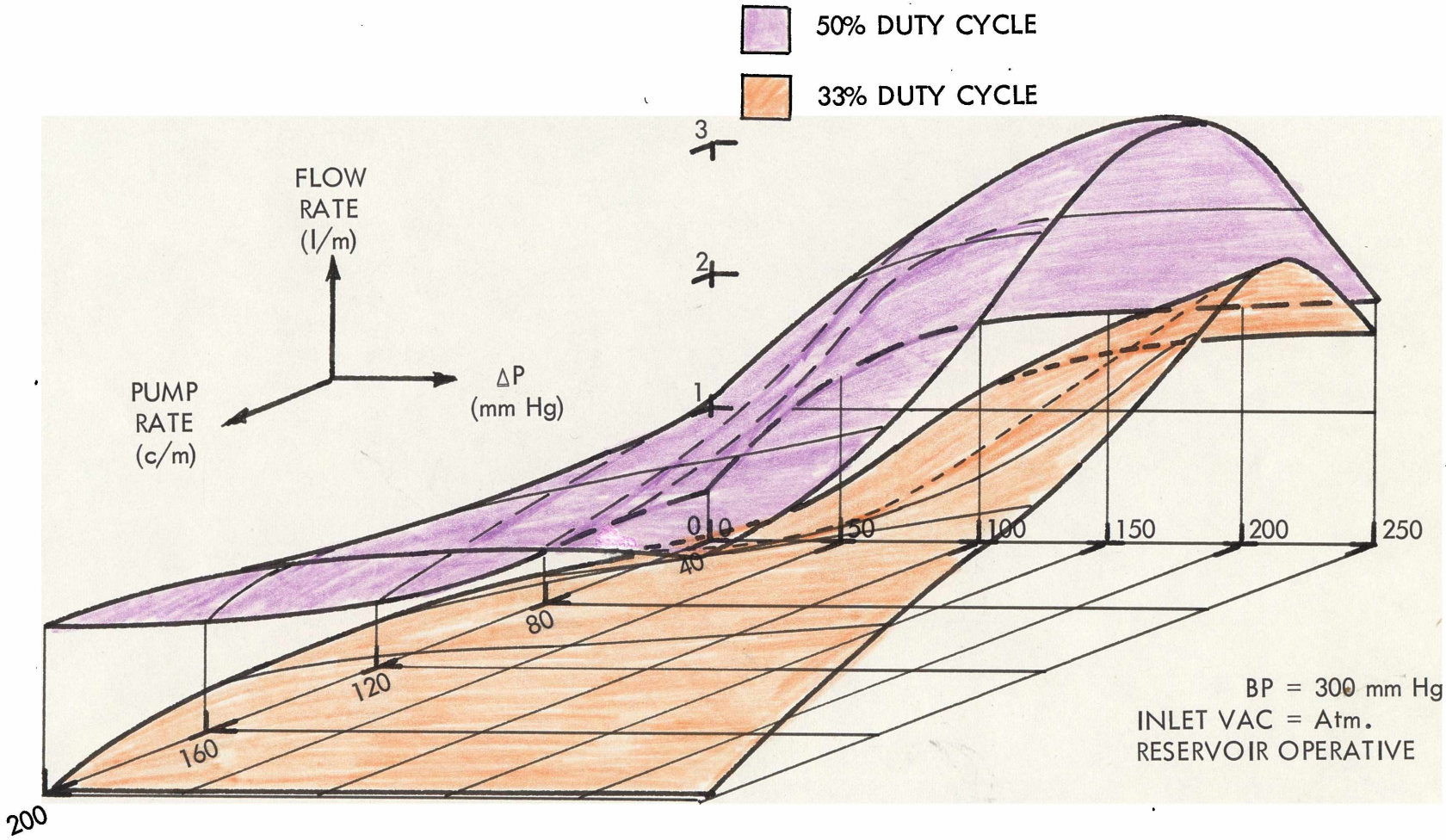


FIGURE 29

FLOW RATE, PRESSURE DIFFERENCE (BAG-DISCHARGE), PUMP RATE SURFACES

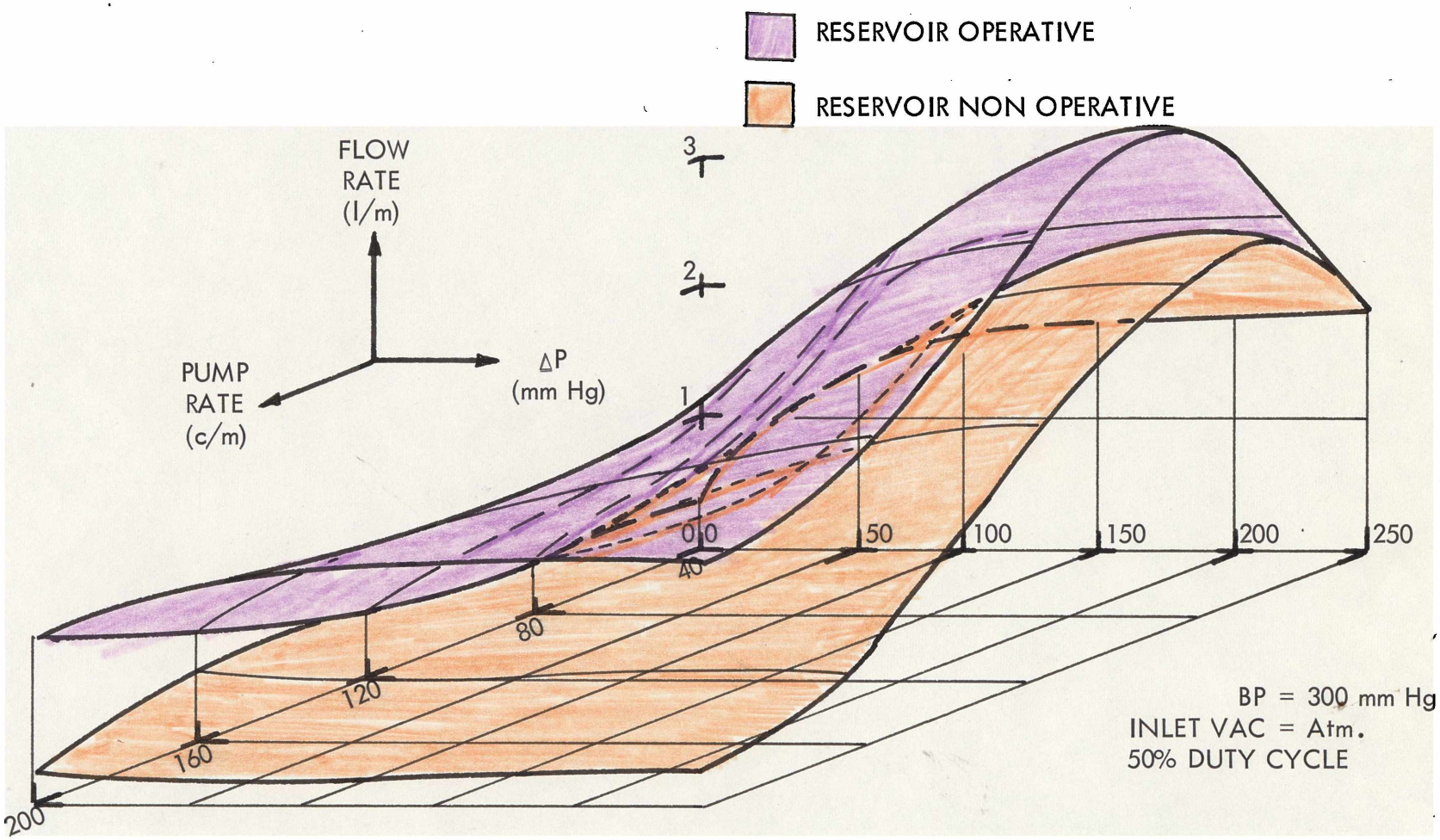


FIGURE 30

FLOW RATE, PRESSURE DIFFERENCE (BAG-DISCHARGE), PUMP RATE SURFACES

In the former, a bag pressure of 300 mm Hg and a discharge pressure of 100 mm Hg were used as typical values. The latter axes maintained 300 mm Hg bag pressure and atmospheric inlet conditions. To facilitate gaining quantitative information, a few typical two dimensional plots of these pump surfaces are shown in Figures 31, 32, 33 and 34.

As a comparison, a pump consisting of two of the smaller pump chambers operated in the parallel, push-pull fashion was tested using the same experimental procedure. The results of these tests are shown in Figures 35 and 36, where comparison is made with the reservoir pump. To complete the picture, both new pumps are compared with the original push-pull pump in Figures 37 and 38. Again constant values of 300 mm Hg bag pressure, 100 mm Hg discharge pressure and atmospheric pressure at the inlet were maintained where applicable. Further clarification of the data is given in the two dimensional plots at Figures 39, 40, 41, 42 and 43.

Results of Flow Studies

These data yield much information concerning the operating characteristics of all three pumps. A summary of the important parts of this information is given below.

Maximum flow rates with the reservoir and new parallel pumps were 3.9 and 5.0 liters per minute at 90 and 120 cycles per minute, respectively, as compared to 6.2 liters per minute at 65 cycles per minute for the old parallel pump with its larger bags. In all cases the maximum flow was achieved with

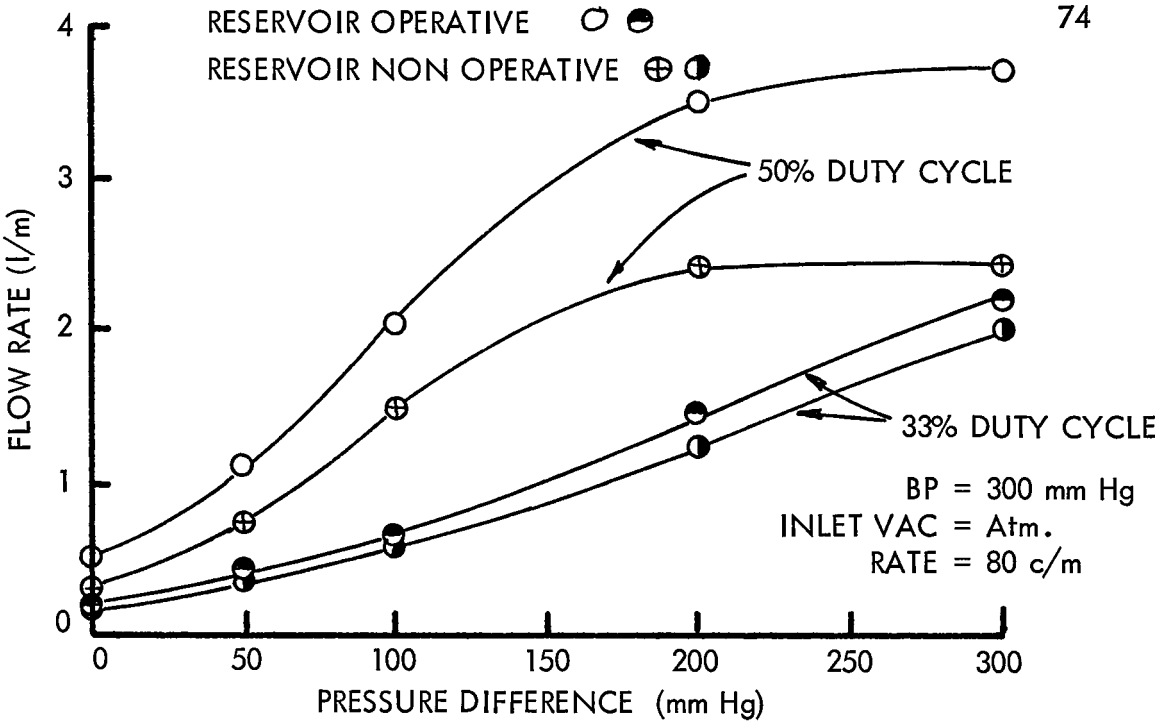


FIGURE 31

FLOW RATE VERSUS PRESSURE DIFFERENCE (BAG-DISCHARGE)

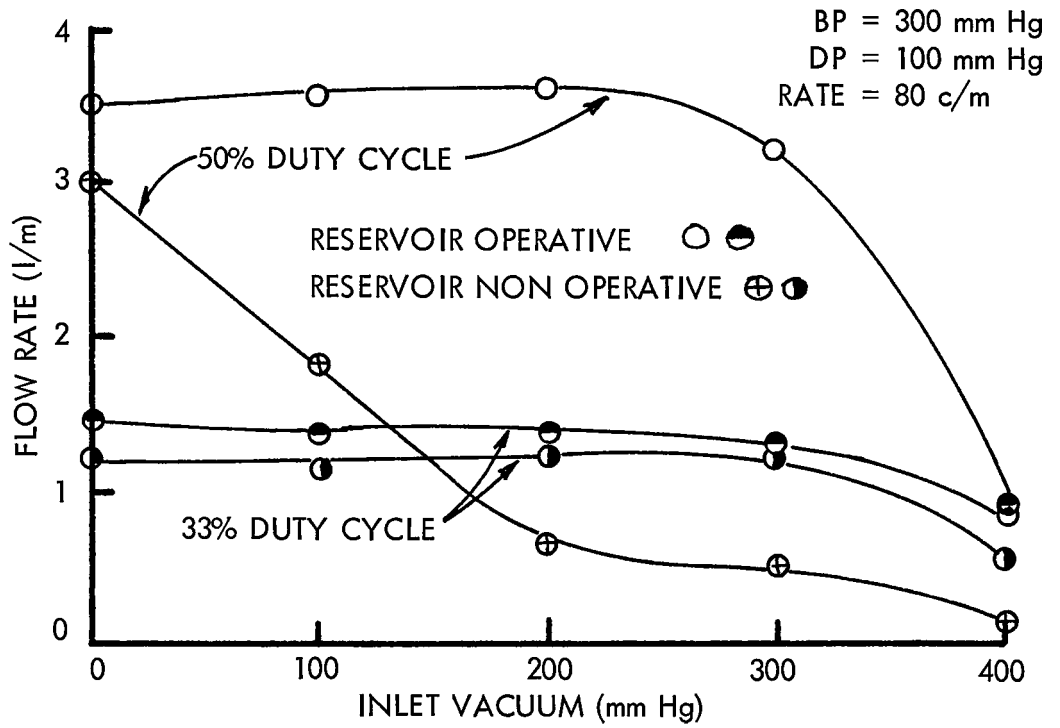


FIGURE 32

FLOW RATE VERSUS INLET VACUUM

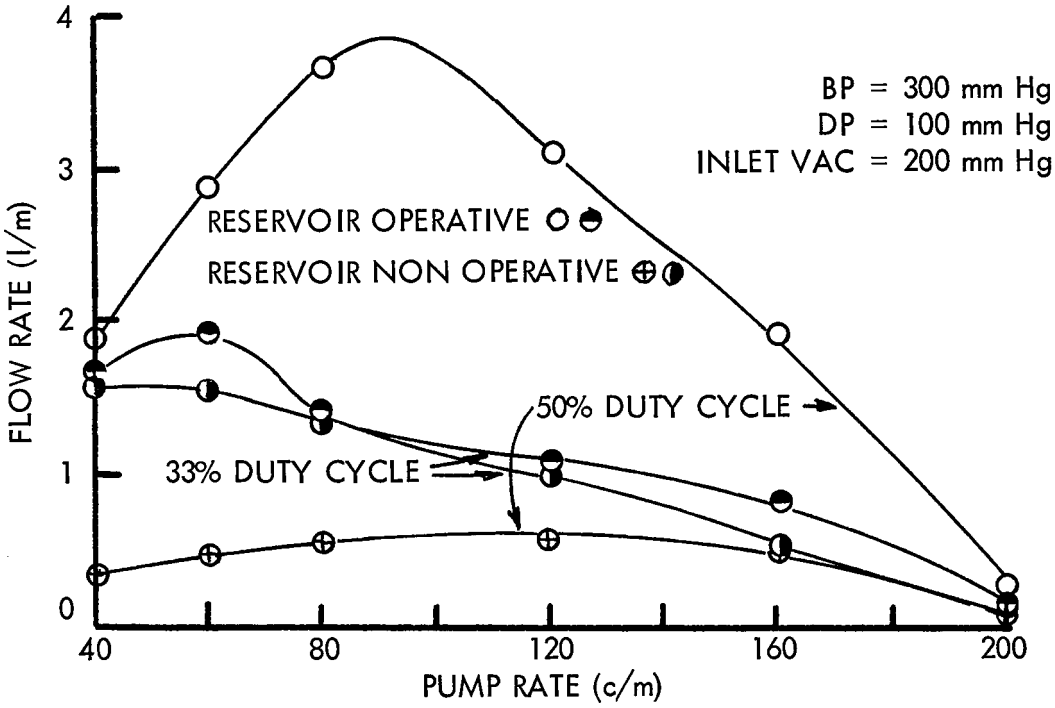


FIGURE 33

FLOW RATE VERSUS PUMP RATE

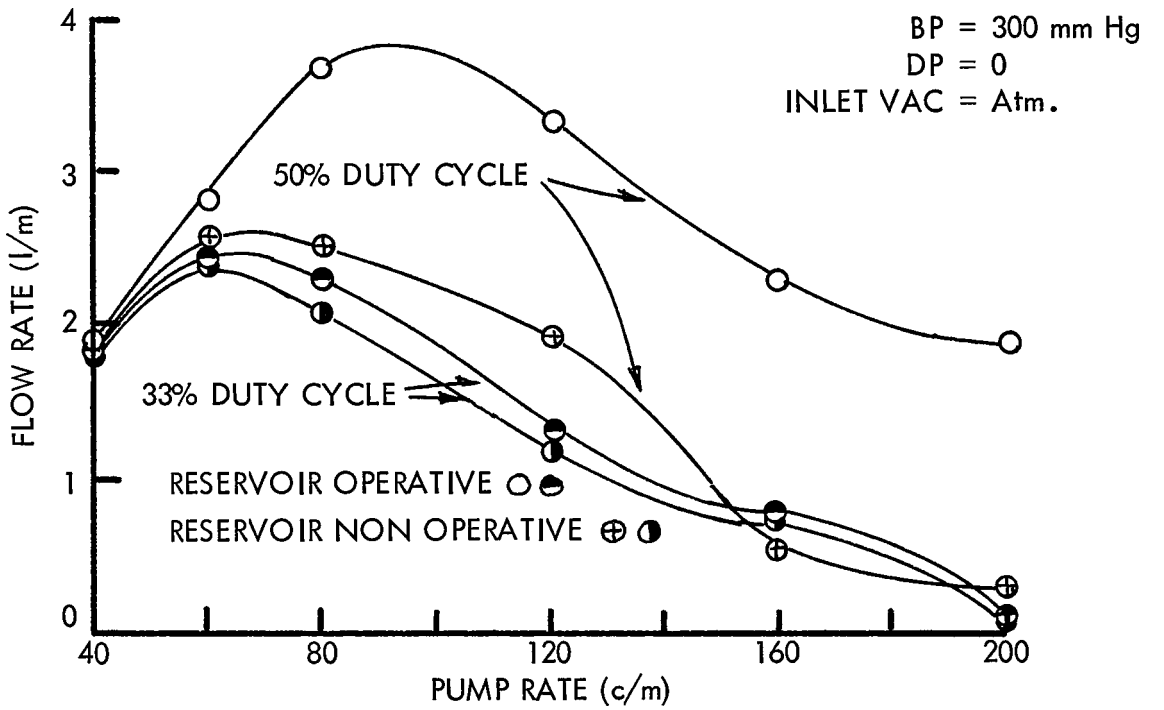


FIGURE 34

FLOW RATE VERSUS PUMP RATE

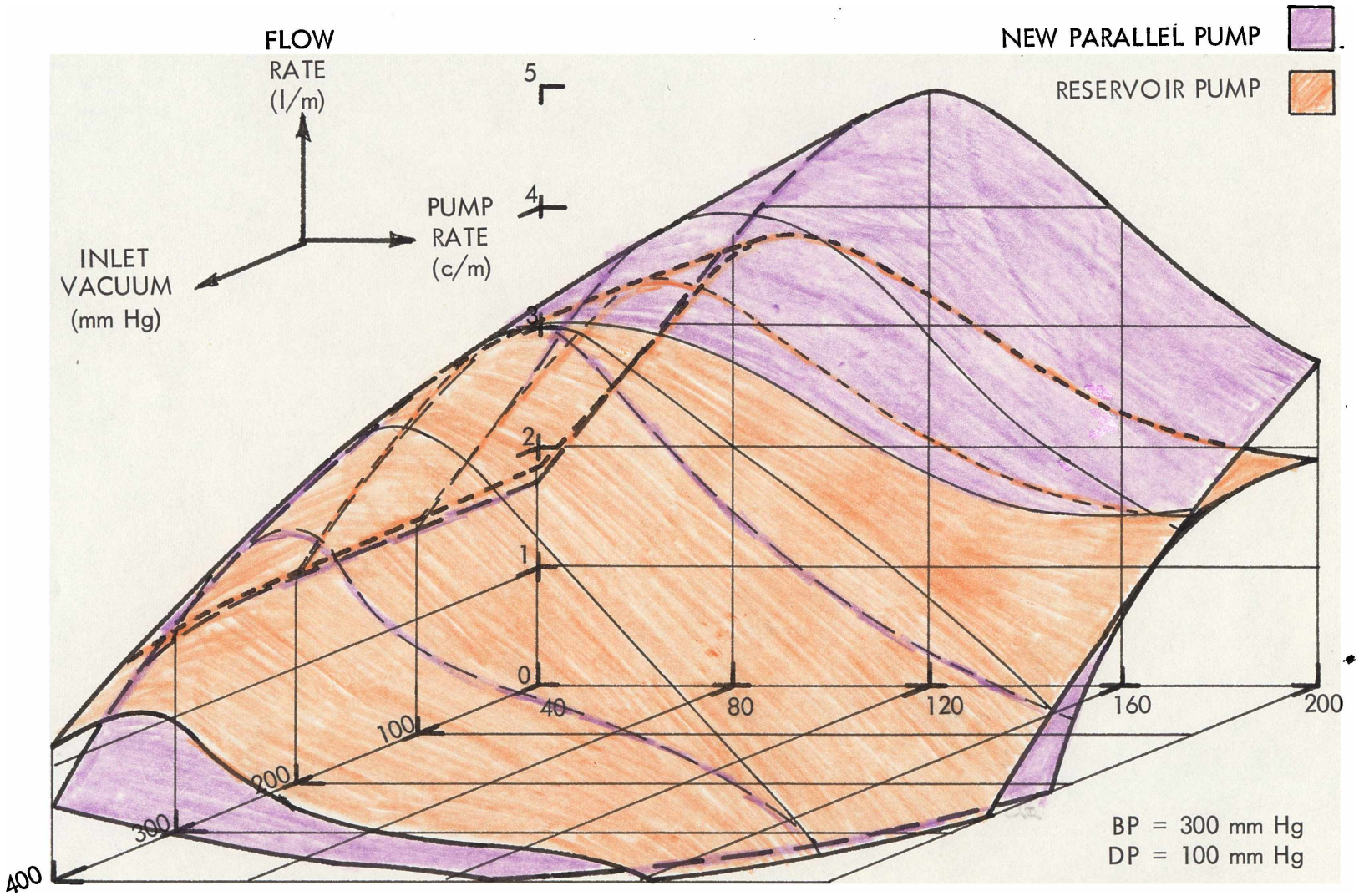


FIGURE 35

FLOW RATE, PUMP RATE, INLET VACUUM SURFACES

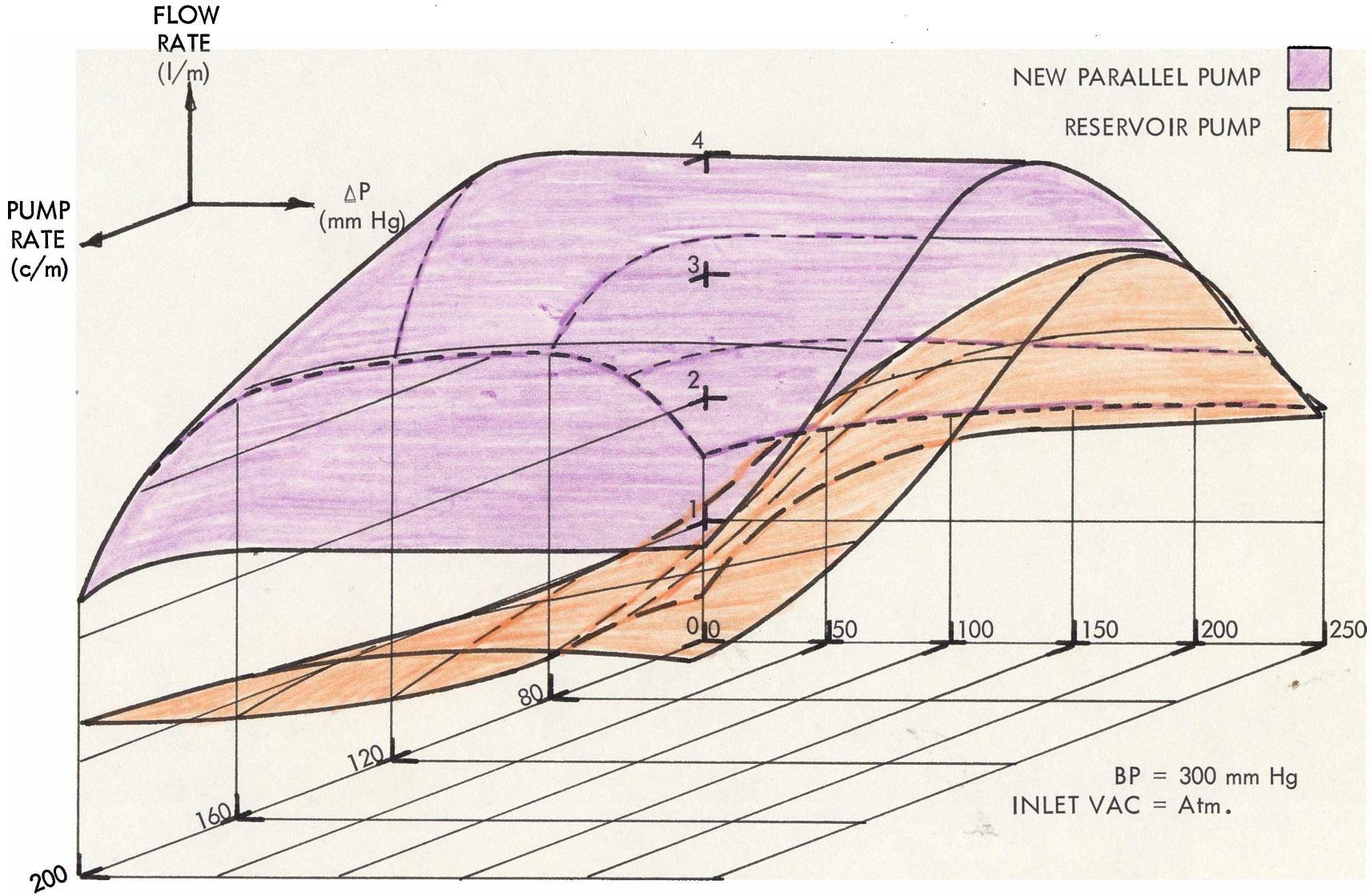


FIGURE 36

FLOW RATE, PRESSURE DIFFERENCE (BAG-DISCHARGE), PUMP RATE SURFACES

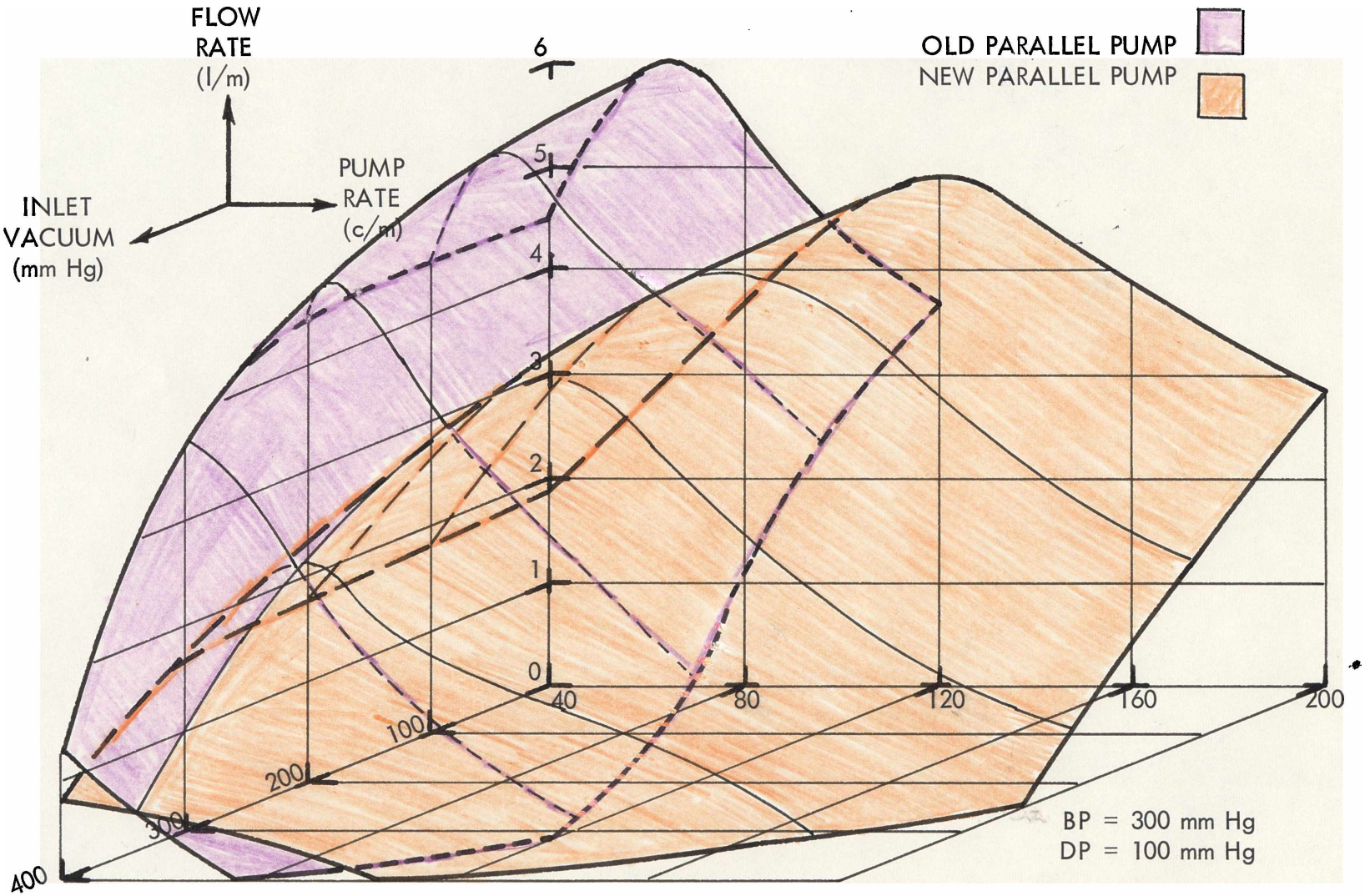


FIGURE 37

FLOW RATE, PUMP RATE, INLET VACUUM SURFACES

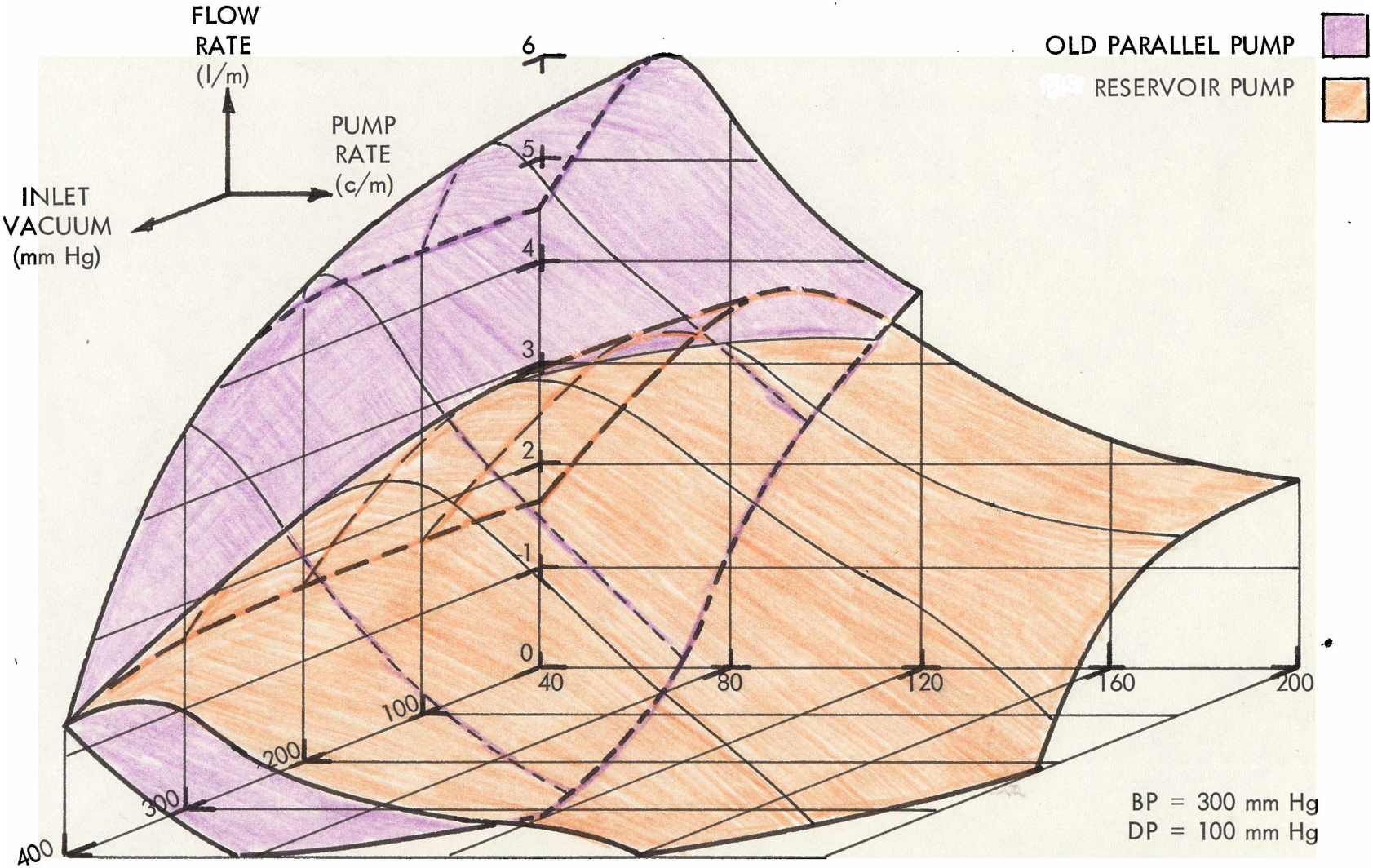


FIGURE 38

FLOW RATE, PUMP RATE, INLET VACUUM SURFACES

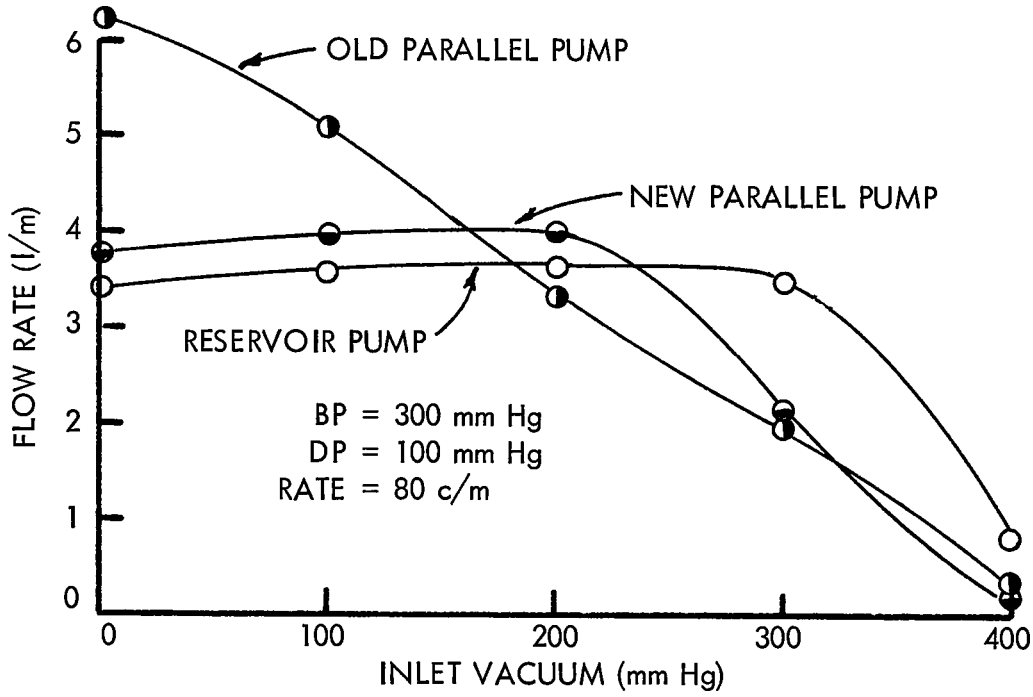


FIGURE 39

FLOW RATE VERSUS INLET VACUUM

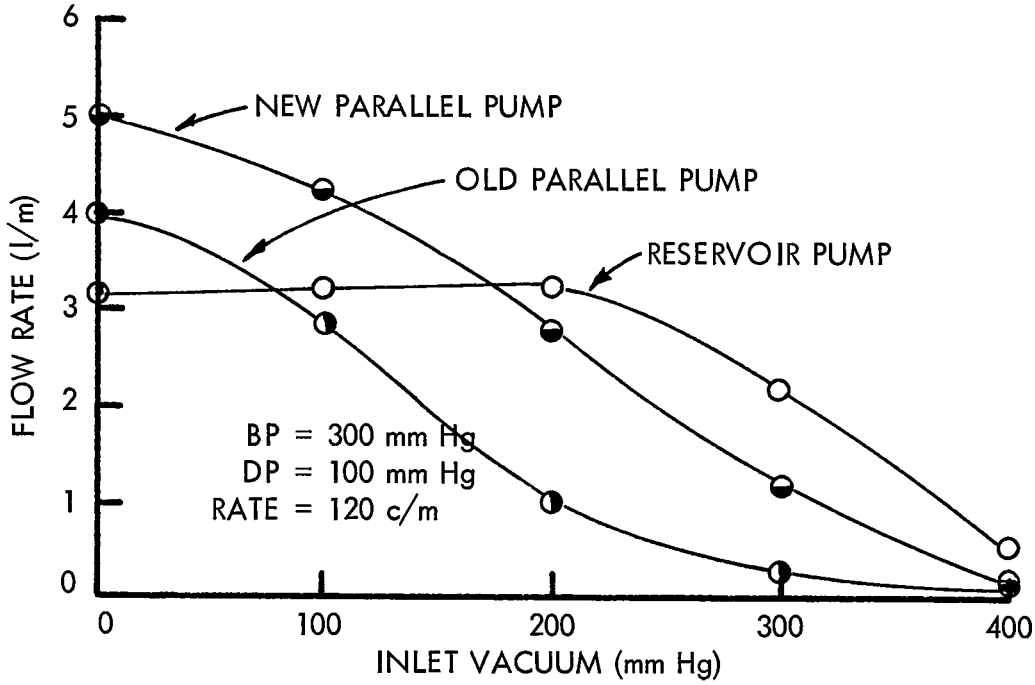


FIGURE 40

FLOW RATE VERSUS INLET VACUUM

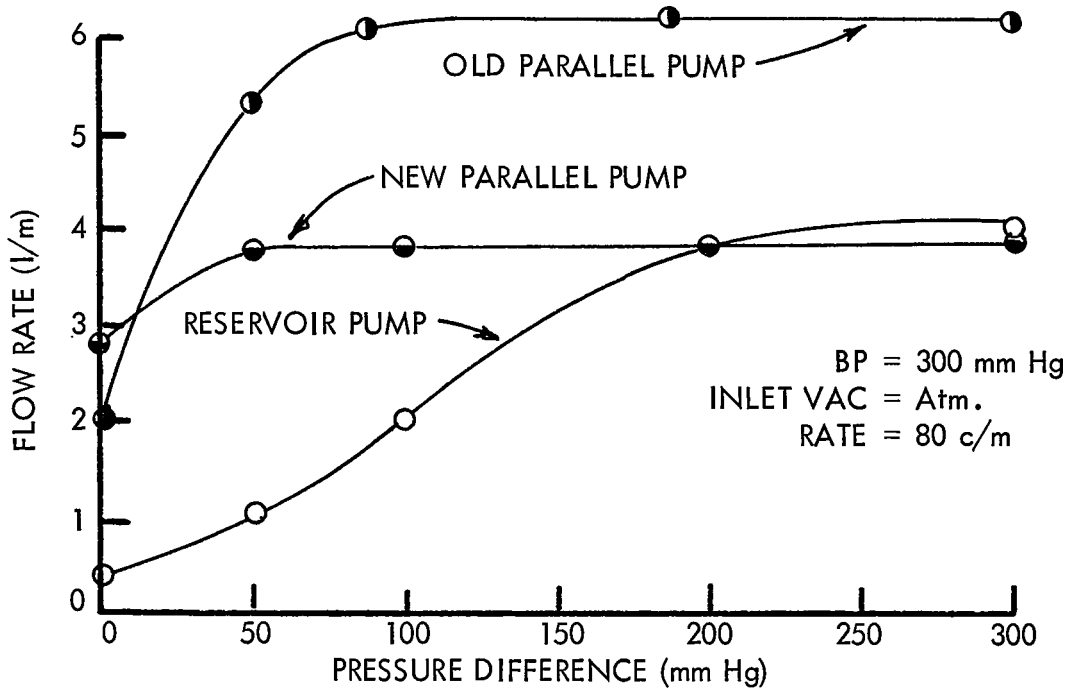


FIGURE 41

FLOW RATE VERSUS PRESSURE DIFFERENCE (BAG-DISCHARGE)

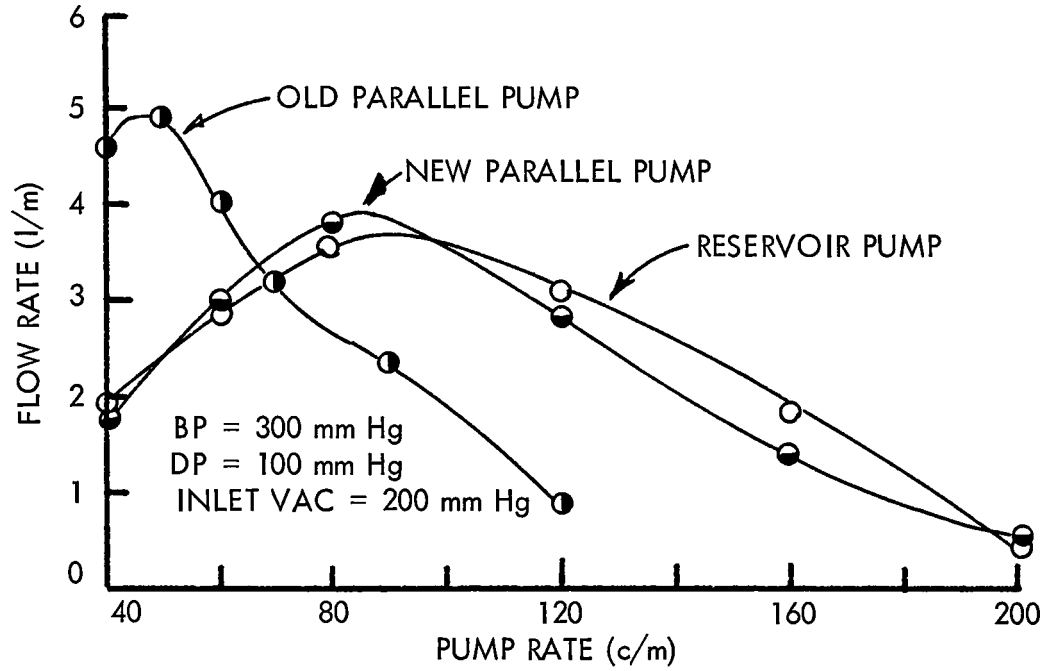


FIGURE 42

FLOW RATE VERSUS PUMP RATE

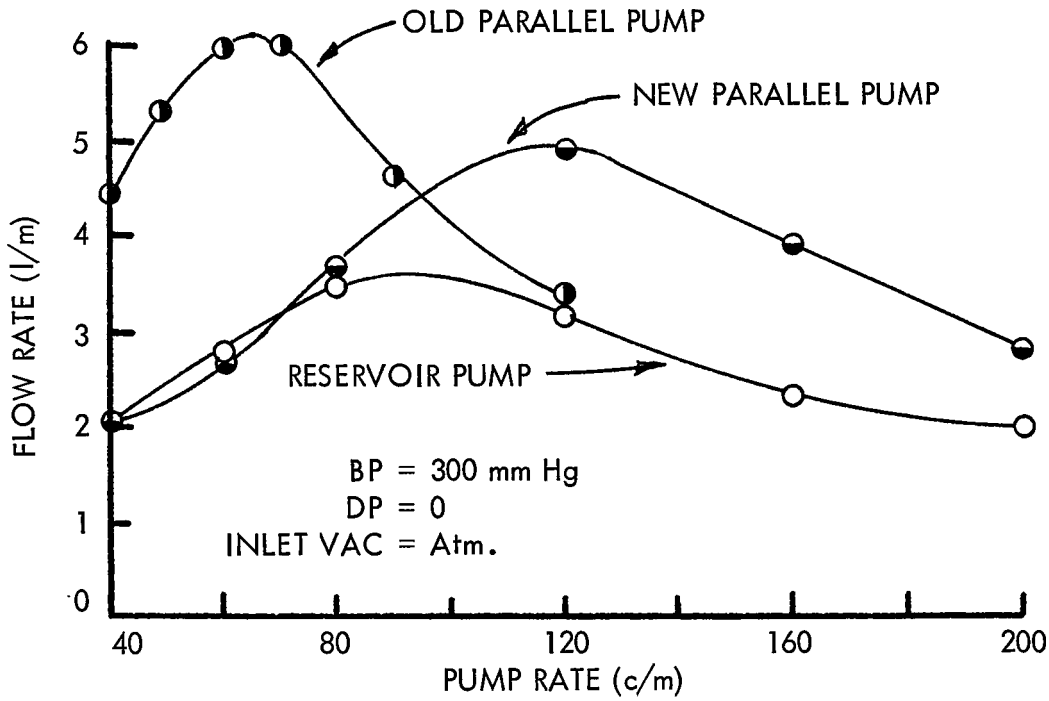


FIGURE 43

FLOW RATE VERSUS PUMP RATE

300 mm Hg bag pressure and zero discharge pressure and inlet vacuum. In the reservoir pump, two questions need to be answered immediately; they are: what effect does the reservoir have on flow rates, and what is the effect of varying the duty cycle of the pumping chamber? The first of these is answered in Figures 26 and 27 where flow rates with and without operative reservoirs are compared for duty cycles of 50% and 33%. In both cases, flow rates were increased for all frequencies and inlet vacuums when the reservoir was operative. This is particularly true in the case of the 50% duty cycle where significant increases were noted. In the case of the 33% duty cycle, increases were present, but of only minimum magnitude. When the 33% and 50% duty cycles with operative reservoir are compared in Figure 28, it is obvious that flow rates were better with the 50% duty cycle. This was true at all frequencies and inlet vacuums with one exception. At frequencies between 80 and 160 and at inlet vacuums greater than 300 mm Hg, a slight edge in flow rate was held by the 33% duty cycle. In addition, as shown in Figure 29, the 50% duty cycle was less sensitive to changes in discharge pressure. Whereas the flow rates with the 33% duty cycle, both with and without the operative reservoir, were relatively independent of inlet vacuum, as shown in Figure 32, their values were low. This same figure shows that the operative reservoir greatly extended the range of inlet vacuums with which good flow rates were maintained with the 50% duty cycle. The curve is nearly flat until an inlet vacuum of 300 mm Hg is reached. Based on these comparisons, only the 50%

duty cycle with operative reservoir will be discussed further.

How does the operation of the reservoir pump compare with the new and old parallel pumps? Comparisons with the new, small parallel pump are shown in Figures 35, 36, 39, 40, 41, 42 and 43, and with the old parallel pump in Figures 38, 39, 40, 41, 42 and 43. Also in Figures 37, 39, 40, 41, 42 and 43 the new parallel pump is compared with the old parallel pump. These figures include Flow Rate vs. Pump Rate vs. Inlet Vacuum Surfaces and Flow Rate vs. Pressure Difference vs. Pump Rate Surfaces and two-dimensional plots taken from these for clarification.

The above-mentioned figures show that the new parallel pump had operating features quite comparable to the old parallel pump with a shift in pump rate. That is, they both operated relatively independent of discharge pressure, until the discharge pressure approached bag pressure, and they both had nearly the same inlet vacuum characteristics. The major difference was in the pump rate for maximum flow which was raised from 65 to 120 cycles per minute.

When the reservoir pump is compared to the two parallel pumps, however, some important differences are noted. In both cases, the flow rates at higher (above 200 mm Hg) inlet vacuums were increased with the reservoir pump. This is even more obvious at higher pump rates, particularly when compared to the old parallel pump. The pump rate for maximum flow with the reservoir pump was 90 cycles per minute. Thus, even though the maximum flow rate for this

pump was not equal to either of the other pumps, it does have sufficient flow to maintain experimental animals and has some characteristics which make it more desirable. As shown in Figure 41, however, it was much more sensitive to changes in discharge pressure than either of the parallel pumps.

Some Comments on Pump Chamber Design

During the process of developing the pump chamber, several changes were made in the design. The pump chamber is shown in Figure 23 and the final Silastic bag design is shown in Figure 25. With one exception the design of the Lucite chambers was straightforward. Originally, the air inlet lines to the bag and valves were not relieved inside the Lucite chambers. However, at times the Silastic was drawn up against the hole and occluded it. The obvious result was loss of vacuum to the chamber. This problem was solved by relieving both the valve and bag chambers around the air inlet hole.

The final design of the Silastic valve and bag assembly was not quite so straightforward. Original designs had several problems, including sealing difficulties, fatigue failures, excess material excursions and stresses. The bevel-squeeze type seal on the valve ends has been quite effective and easy to assemble. Originally, it was hoped to constrain the bag ends to keep them from collapsing, but most attempts failed. Finally, a perforated metal stabilizing plate was included at one end of the bag, and the other end was stabilized by the addition of a small silicone adhesive sealant (Clear Seal, General Electric). The perforated plate serves not only as a restrainer for the bag ends,

but also as a stiffener to reduce excess motion of the membrane between the lower valve and bag chambers. Since these two chambers are pressurized and evacuated out of phase, a large pressure difference is developed across the membrane. Without extra stiffening, this resulted in several fatigue failures of the Silastic and silicone materials. The design shown in the figure has had many hours of trouble-free operation.

In Vitro Hemolysis Studies

Utilizing the same experimental procedure and apparatus as described previously for the parallel bag push-pull pump provided for the undertaking of a series of in vitro hemolysis studies on the reservoir pump. The measurements made included hemolysis rates with and without restricted inlets to the pump. All tests, with one exception, were 6 hours in duration and similar flow rates (3.2 l/m) were maintained. The pump frequency was held at 90 cycles per minute because this was the frequency for maximum flow.

The results of these experiments are shown graphically as plasma hemoglobin level versus time in Figure 44, and summarized in Table VI. In addition, these data are compared with the values of hemolysis from previous tests and in vivo experiments in Figure 45. These data show average hemolysis rates for restricted inlet, non-occlusive bag operation in the reservoir pump of 80 mg Hb liberated per hour. This compares with an average hemolysis rate of 71 mg Hb liberated per hour with non-restricted inlet, non-occlusive operation. Again, as previously obtained with the parallel pump,

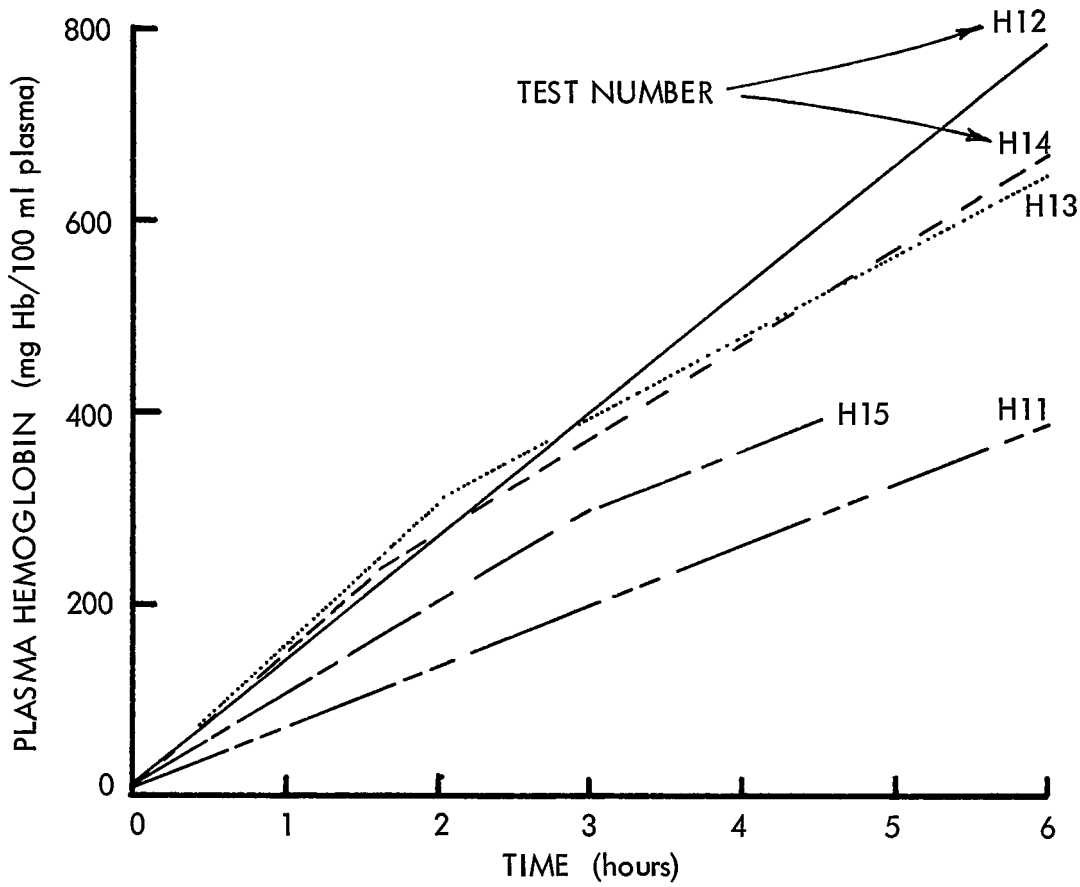


FIGURE 44

PLASMA HEMOGLOBIN VERSUS TIME, RESERVOIR PUMP

TABLE VI

IN VITRO HEMOLYSIS STUDIES II

Test No.	Pump Type	Inlet Vacuum (mm Hg)	Bag Condition	Test Duration (hours)	Flow Rate (l/m)	mgms Hb Liberated per hour*	Index of Hemolysis	Blood Hct.
H11	Reservoir	0	Open	6	3.2	44	0.20	20
H12	Reservoir	150	Open	6	3.2	87	0.40	20
H13	Reservoir	200	Open	6	3.2	138 (0-2 hrs) 73 (2-6 hrs)	0.64 0.34	19 18
H14	Reservoir	0	Open	6	3.2	121 (0-1½ hrs) 87 (1½-6 hrs)	0.53 0.39	23 22
H15	Reservoir	0	Open	4 1/2	3.2	83 (0-3 hrs) 48 (3-4 hrs)	0.40 0.24	15 14

Pump Rate 90 c/m

* Volume corrected to previous hemolysis studies by multiplying actual rate by ratio $\frac{\text{Test Volume}}{\text{Previous Test Volume}}$

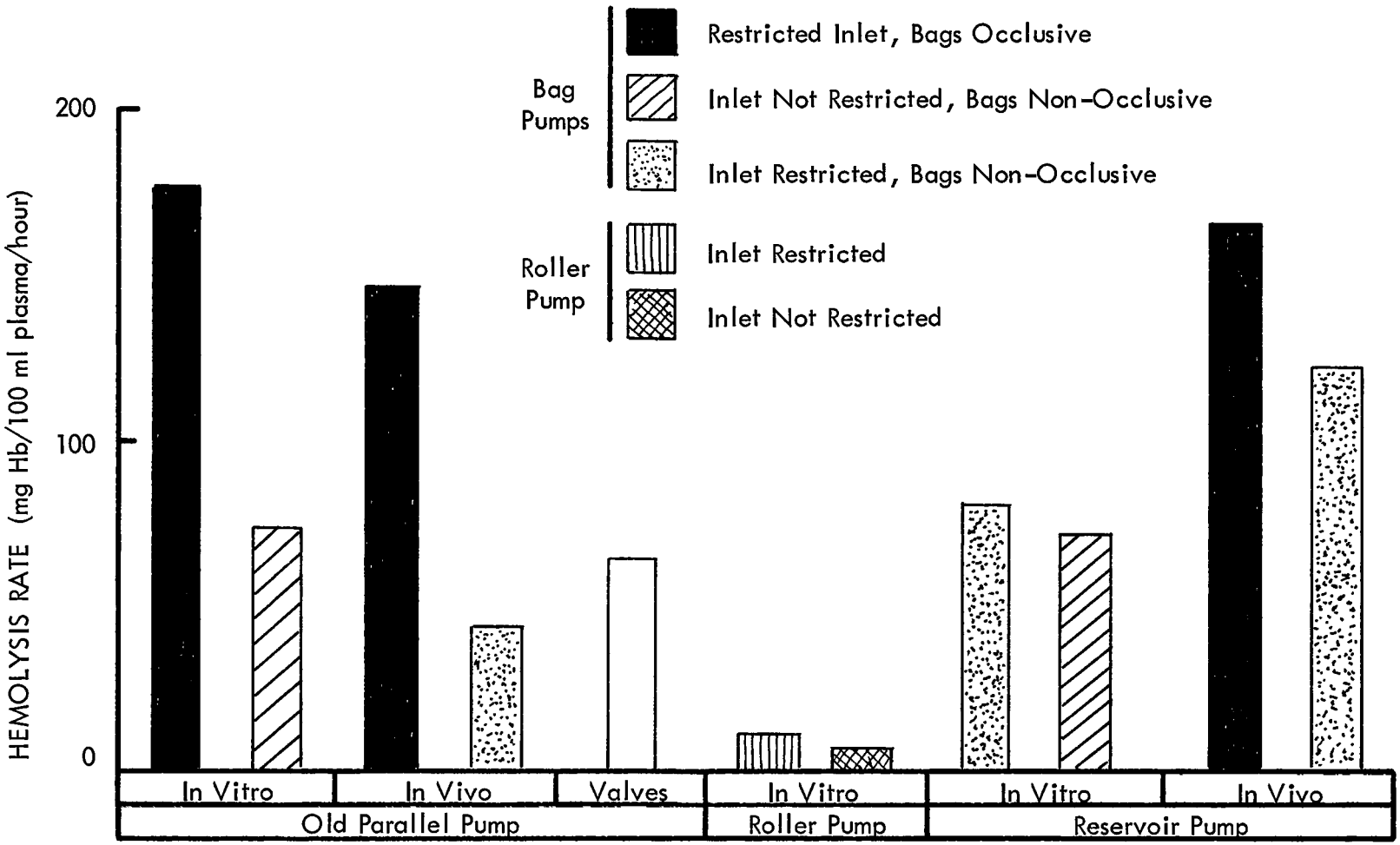


FIGURE 45

HEMOLYSIS STUDIES SUMMARY

the slight difference in hemolysis rates indicates that the major source of hemolysis was the occlusive valves rather than restrictions of the bypass method. Inlet vacuums as high as 200 mm Hg vacuum did not seem to increase the hemolysis rate appreciably.

Also, in this series of experiments the Index of Hemolysis (IH) was lower than previously obtained. The reason appears to be the increased flow rate available from the reservoir pump. The flow rate was higher than for the other tests, further indicating that the hemolysis rate for this type of pump was a function of variables other than flow rate. In fact, the corrected hemolysis rate for the valves was 50 mg Hb per hour, which compares with an average liberation rate for the reservoir pump of 75 mg Hb per hour. The rate-correction is made by multiplying the liberation rate of one old pumping chamber by the ratio of the pumping rates (i.e., 90/66 = 1.37). Thus, the valve hemoglobin-liberation rate accounts for the major portion of the reservoir pump hemolysis when operated in a non-occlusive manner.

In Vivo Pump Studies

In order to test the effectiveness of the reservoir pump in maintaining total circulation in living animals, a series of in vivo studies was carried out. As before with the parallel pump, dogs weighing from 18 to 25 kg were anesthetized with sodium pentobarbital (30 mg/kg) and proper catheters installed for pumping by means of the Geddes-Schuhmann method of left ventricular

bypass. After the pump was turned on, ventricular fibrillation was induced to insure total circulatory maintenance. When possible, after a period of pumping, the heart was electrically defibrillated and recovery monitored. A summary of these studies appears in Table VII.

The first test was successful in that pump flow rates were high and, after an hour of pumping, an arterial blood pressure of 125/80 was obtained. In addition, the blood pressure waveform was near physiological as shown in Figures 46 and 47. As a comparison, a normal Physiograph record of a normal dog before pumping is shown in Figure 48. For the first time since the beginning of collapsible-bag pump studies, pump inflow from the animal could be maintained with less difficulty than pump outflow, even with a bag collapse pressure of 400 mm Hg. This was due to the large pressure drop across the femoral artery catheter. Unfortunately, hemolysis rates were quite high, even though the bags remained open. These data, however, were questionable due to fibrin collection in the blood samples.

In the remaining tests an additional femoral artery was catheterized for infusion of blood from the pump. This facilitated the discharge of blood from the pump at reasonable bag collapse pressures. This change also resulted in more physiologically appearing arterial pulse waveform.

The second test results were not so encouraging, because adequate flow could not be maintained with ventricular fibrillation. Pump flows with the heart beating were high and, in fact, responsible for most of the total blood

TABLE VII

IN VIVO HEMOLYSIS STUDIES II

Test No.	Bag Condition	Pumping Duration (hours)	Flow Rate (l/m)	Blood Hct.	mgms Hb Liberated per hour	mgms Hb Cleared per hour	Remarks
82-RP1	Open	1 1/2	1.4	52	244 (147)*	140 (initial) 12 (later)	Nice Pulse Form, Good Recovery
83-RP2	Closed Closed	2/3 2/3	Low Low	36 24	165 180	50 --	Nice Pulse, Flow good before fibrillation; poor following.
84-RP3	Part Open	3 1/4	1.8-0.4	38	97	--	Initial good flow, poor later; Spontaneous fibrillation after pumping.
85-RP4	Closed	1 3/4	3.3-0.4	36	144	--	Low Flow, Spontaneous fibrillation after pumping.

Pump Rate 90 c/m

* A more probable value due to fibrin problems in blood samples.

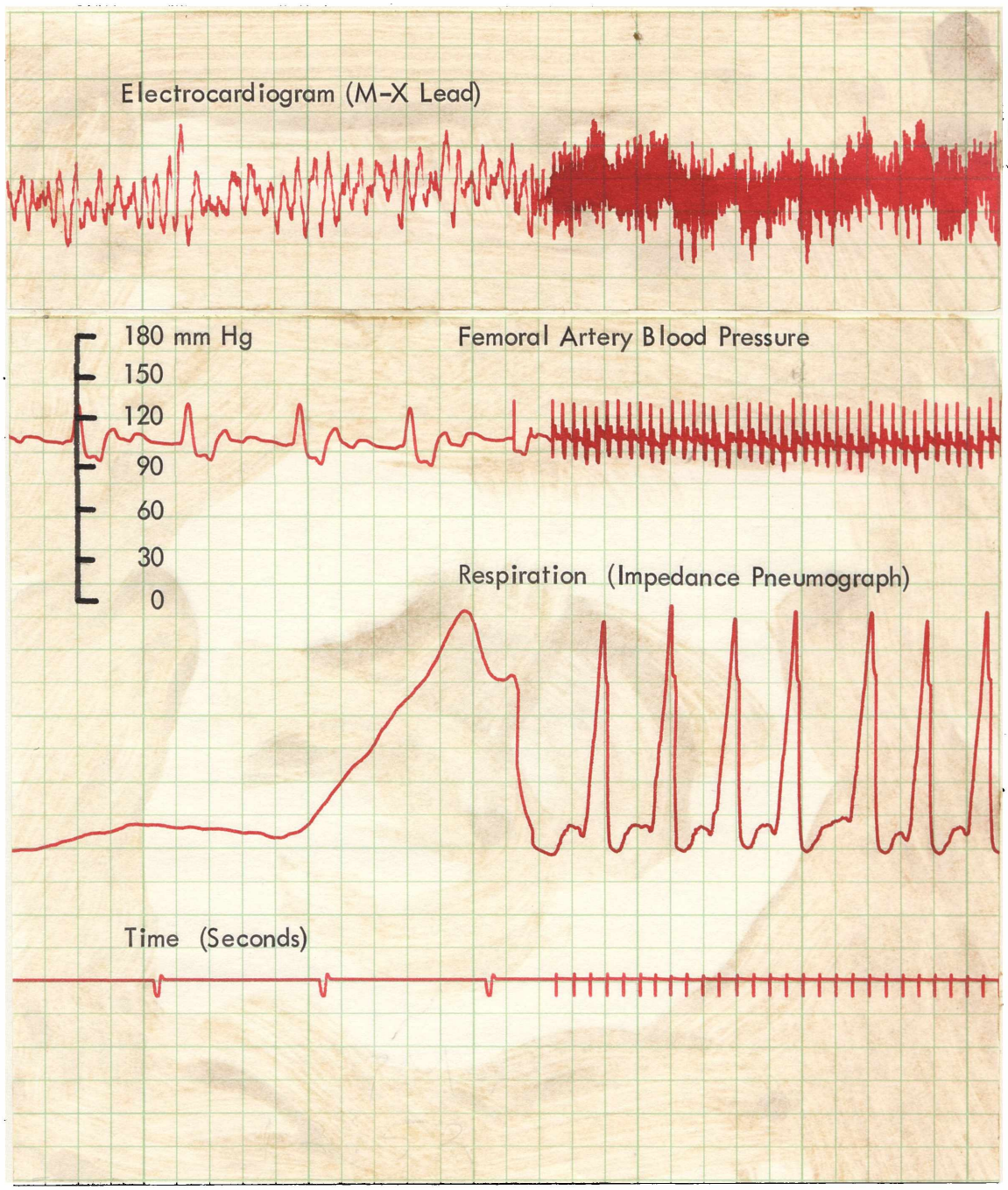


FIGURE 46

SAMPLE PUMPING PHYSIOGRAPH RECORD, EXPT. 82-RP1

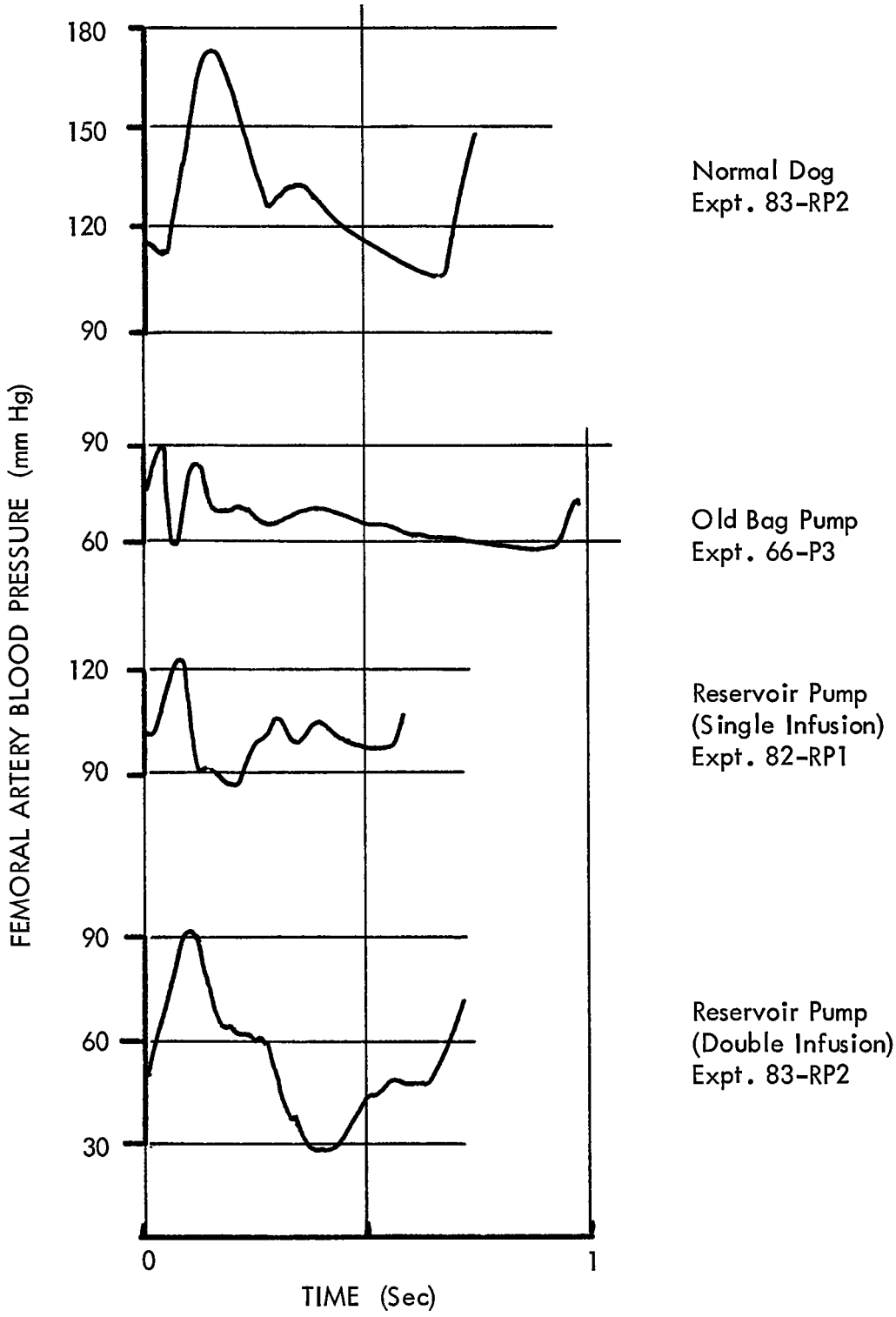


FIGURE 47

PULSE WAVEFORM SUMMARY

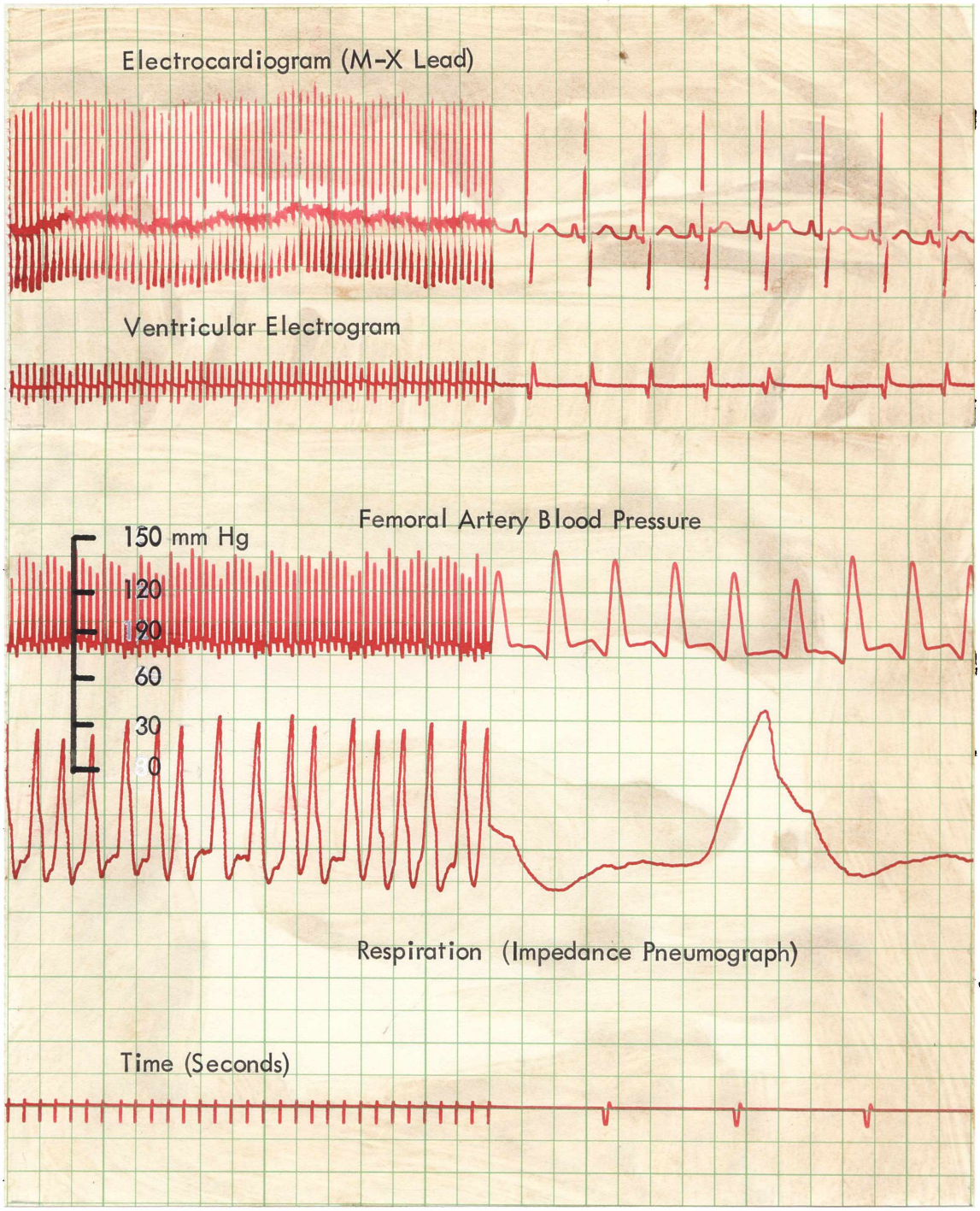


FIGURE 48

SAMPLE NORMAL PHYSIOGRAPH RECORD, EXPT. 66-P3

flow as evidenced by the total absence of pressure pulse waves produced by the heart. As soon as fibrillation was induced, flow immediately dropped to negligible amounts. This problem has been mentioned before and is now attributed to collapse of the ventricles around the catheter tip, possibly due to high pulmonary flow resistance and/or peripheral blood pooling. Due to this problem, the pump bags operated occlusively and the resulting hemolysis rates were high.

After 2/3 hour of pumping, due to the low arterial blood pressure, the ventricles were electrically defibrillated. Following removal from the pump, the animal developed a state of shock or progressive congestive failure, as evidenced by a continuing drop in blood pressure. An intravenous drip could not reverse the situation and the pump was started again. Pump flow and arterial blood pressure were high until fibrillation was again induced. Since adequate flow could not be maintained the experiment was then terminated. A sample Physiograph record is shown in Figure 49.

The third and fourth in vivo experiments with the reservoir pump were almost identical. In both cases, flow rates through the pump before and a short time after fibrillation were good (1.8 l/min and 3.3 l/min). Soon after fibrillation was induced, however, flow rates dropped, pump inlet vacuums increased to near 400 mm Hg vacuum and arterial pressure dropped. The animals were maintained at this low level of circulatory maintenance for approximately one hour before defibrillation was attempted. All during the time of pumping, vital signs, such as tendon jerk reflexes, remained good. Initial defibrillation was

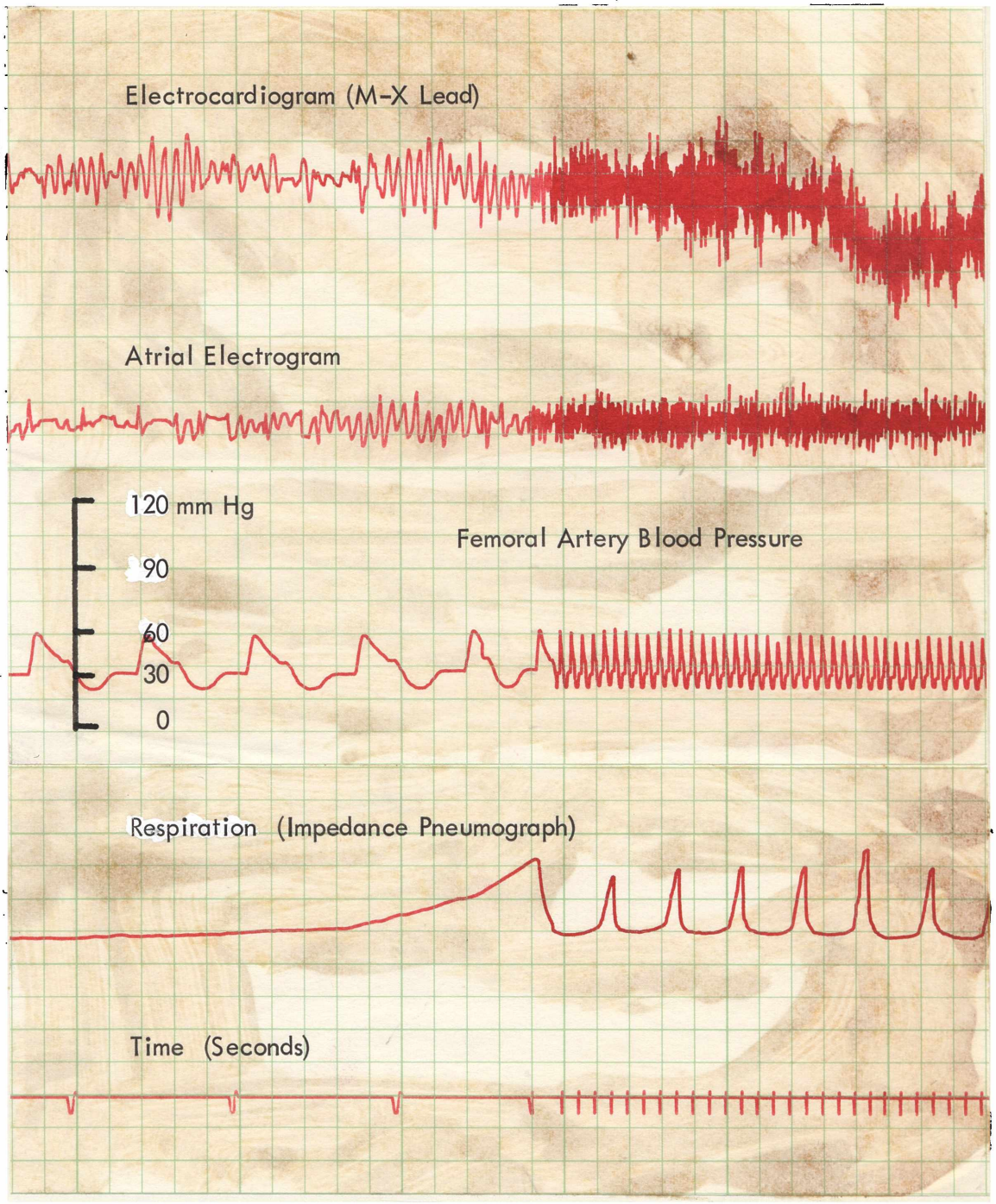


FIGURE 49

SAMPLE PUMPING PHYSIOGRAPH RECORD, EXPT. 83-RP2

difficult and required several attempts before success. After a short period of normal heart function, the animal was weaned from the pump. Soon thereafter spontaneous fibrillation occurred and defibrillation was again attempted. This sequence was repeated several times in both animals until termination of the experiment without successful long term recovery. In both cases circulation was maintained without difficulty by the reservoir pump during the defibrillation attempts. Total time of circulatory maintenance was 3 1/4 and 1 3/4 hours for the third and fourth experiments respectively.

One difference was noted in the third experiment in that initially during the pump run, following fibrillation, flow rate was good (1.8 l/min), inlet vacuum low (150 mm Hg vacuum) and arterial pressure was rising. In an attempt to increase the rate of arterial pressure rise, the bag collapse pressure was increased. The almost immediate result was increasing inlet vacuum, decreasing arterial pressure and flow rate; a trend that could not be reversed. The problem in these experiments was again attributed to collapse of the myocardium around the catheter tip. The results of experiment number three indicate the existence of a maximum flow, over which the pulmonary circuit or other parameters were unable to handle and the blood return to the left ventricle was less than that being delivered to the systemic arteries.

Since, in the fourth test, the bag remained collapsed at all times the resultant hemolysis rate was high. During the third test the bags remained mostly open and the hemolysis rates were somewhat lower as shown in Table VII.

Comments on Test Results

Even though the results of these experiments were not as successful as hoped, in that flow rates were low during fibrillation, it was shown that the reservoir pump was capable of total circulatory maintenance in dogs for extended periods of time. The low flow problem was attributed to parameters of the left ventricular bypass method rather than problems with the pump itself. Pulse waveforms with adequate flow rates were near physiological. The pump has proved to be easy to operate and quite reliable with no failures of any kind during the course of the in vivo experiments. It should be stated that the tests with ventricular fibrillation are an extremely severe test of the pumping system and not characteristic of cardiac assistance techniques.

Hemolysis rates in these in vivo studies were higher than expected. Most of the blood trauma can be attributed to the occlusive bag and valves. However, it was believed that this was not the only problem because the rates were higher than in the in vitro studies. The reason was believed to be the excessive inlet vacuums previously mentioned. As stated in Chapter III high vacuum is a cause of blood hemolysis.

Another desirable feature of the reservoir in this pump was noticed as a result of the high vacuums. Blood under high vacuum has been noticed to out-gas considerably, particularly when agitated as in a pump. The result is the formation of bubbles in the flow lines. In the reservoir pump, these bubbles were trapped and remained in the top of the reservoir. In no case have any of these bubbles been passed on to the dog.

CHAPTER V

DISCUSSION AND SUMMARY

The descriptions, data and results presented in the previous chapters have shown the development and evaluation of two types of extracorporeal blood pumps for use with left ventricular bypass without thoracotomy. Both of these pumps operated using the collapsible-bag active-collapsible-valve method of pumping and have been shown to be capable of total circulatory maintenance in dogs. The first model tested consisted of two parallel pumping chambers operated in a push-pull fashion. With this pump, circulatory maintenance was achieved for periods of time up to four hours with good recovery following pumping. This pump was characterized by high hemolysis rates, and only minimal pulsations in arterial blood pressure. It suffered a serious reduction in flow rates at increased inlet vacuums and of pump rates above 70 cycles per minute.

After evaluation of the parallel pump, another pump was fabricated, again utilizing the collapsible bag and valve pumping technique. This pump included a new pumping-chamber design, smaller pumping chamber volume, new valve design, and a vacuum-type reservoir. The vacuum reservoir consisted of a flexible Silastic bag which contained the blood and a controllable vacuum applied to the outside of the bag. The reservoir was physically attached to the inlet of the pumping chamber and, when correctly adjusted, partially collapsed

when the pumping chamber filled from it. The resultant inflow to the reservoir was steady and non-pulsatile. This type flow allowed higher flow rates at reduced flow velocities when compared with pulsatile flows.

Flow studies with the reservoir pump have shown it to have superior inlet vacuum characteristics. That is, at increased inlet vacuums, flow rates were higher than with the parallel pump. The smaller size of the pumping chamber increased the pump rate for maximal flow from 65 to 90 cycles per minute and yielded adequate flow rates at pump rates as high as 150 cycles per minute. In addition, arterial pulse in animals showed near physiological waveform and better blood pressure was maintained during circulatory maintenance. In vitro hemolysis studies showed comparable hemolysis rates, even at the increased flow rates, and the resultant lower values of the Index of Hemolysis. The pump was easy to place into operation and, once properly adjusted, required a minimum of attention.

The hemolysis rates, unfortunately, remained higher than desirable. This hemolysis has been shown to be primarily associated with two parameters. First, the bag occlusiveness of the old pump contributed greatly to the red blood cell damage. This problem was partially solved in the reservoir pump since the bag occlusiveness could be partially controlled by not allowing total collapse of the bag. Secondly, the active, collapsible valves were a factor in the hemolysis rates. Even though the valves were redesigned and the occlusive area reduced, relatively high hemolysis rates continued. It appears that this damage

is characteristic of this type valve and very little, if anything, can be done to eliminate totally the high hemolysis rates. The damage varied directly with valve collapse pressure, but reducing this pressure resulted in incompetent valves, and equally distressing problem.

What, then, has this work produced? To begin with, a pump has been developed that can adequately maintain total circulation in dogs with ventricular fibrillation weighing approximately 20 kg for extended periods of time. The pump, when used with the Geddes-Schuhmann method of left ventricular bypass, produces blood pressure waves with nearly physiological shape and magnitude. The pumping chamber is simple and easy to fabricate and has proved to be highly reliable. The electronic circuitry can be arranged so that the pumping cycle can be triggered by the animal's own electrical impulses for further testing in circulatory assist. Operation of the pump has been trouble-free and maintenance was minimal.

Based on the results of these tests, it appears that a pump of this general design should have value in the clinical application of the Geddes-Schuhmann method of left ventricular assist. Whereas the basic design is sound, some refinements and improvements are indicated. These refinements are hemolysis rate reduction and possibly refined fabrication techniques for the Silastic pumping chamber. The present design has a few small surface irregularities which could be sources of thrombus formation and elimination would be desirable.

Areas for Future Work

Future developmental work on this pump should be centered in the following areas.

Hemolysis Reduction: In this investigator's opinion, a major portion of the hemolysis could be reduced by replacing the active, collapsible valves with one of the more common passive valve types. In particular the ball-in-cage prosthetic heart valve should be well-suited for this purpose. Hemolysis rates for these valves are low and their use has been clinically proved.

Thrombus Formation: By using different fabrication techniques for the collapsible bags, reduction in the formation of thrombo-emboli should result. By the use of more sophisticated materials, such as the Dacron Velour linings, this problem should be virtually eliminated.

Control Circuitry and Pneumatic System: Increasing the size of the pneumatic tubing and reservoir tanks should result in increased frequency response of the pump. Also, the addition of electronic circuitry to facilitate variations in duty cycle would be desirable to gain additional control over bag occlusiveness. Repackaging of the control circuitry and pneumatic system could also be done to make it more esthetically pleasing and also to reduce the ambient noise level from the solenoid valves and vacuum pumps.

Flow Rate Monitor: Even though the present pump output transducer operates satisfactorily, some time is involved in calculation before actual flow rates are known. It is often desirable to know at once what the actual flow

rate is and the addition of such a device, or method, would be advantageous.

Decreased Flow Rates: This may not be a problem with the pump alone as it could also be associated with the bypass method itself. The actual reason behind these low flow rates with ventricular fibrillation needs to be determined exactly so that steps can be taken to correct the problem. In all probability, the resultant improvements will be refinements in both the method and the pump.

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APPENDIX A

DESCRIPTION OF GEDDES-SCHUHMANN METHOD OF LEFT VENTRICULAR BYPASS

The method of left ventricular bypass of interest to this study is one developed by R. E. Schuhmann, L. A. Geddes and others at Baylor University College of Medicine, Department of Physiology, Houston, Texas. It is the Ph.D. dissertation of one of the developers (R.E.S.). The method has been described in three publications [13, 38, 86], one of which (the latter) includes a description of the first model of the pump developed by this investigator. Since the pump in question was developed for use with this method of left ventricular bypass, it seems appropriate to describe briefly the mechanics of this system.

Basically, the system involves catheterization of the left ventricle to obtain blood for the inlet of the pump; return of the blood to the subject is via femoral and carotid artery cannulae. Entrance to the left ventricle is gained via the right carotid artery, aortic arch and then through the aortic valve (during the opening phase) into the left ventricle. The reinfusion is achieved through a T-tube in the left femoral artery and a cannula in the right carotid artery facing the brain. (See Figure 50 .) It is felt by us and at least one other investigator [29] that this latter catheter maintains better flow to the brain and is therefore desirable. Care must be taken, however, that excess

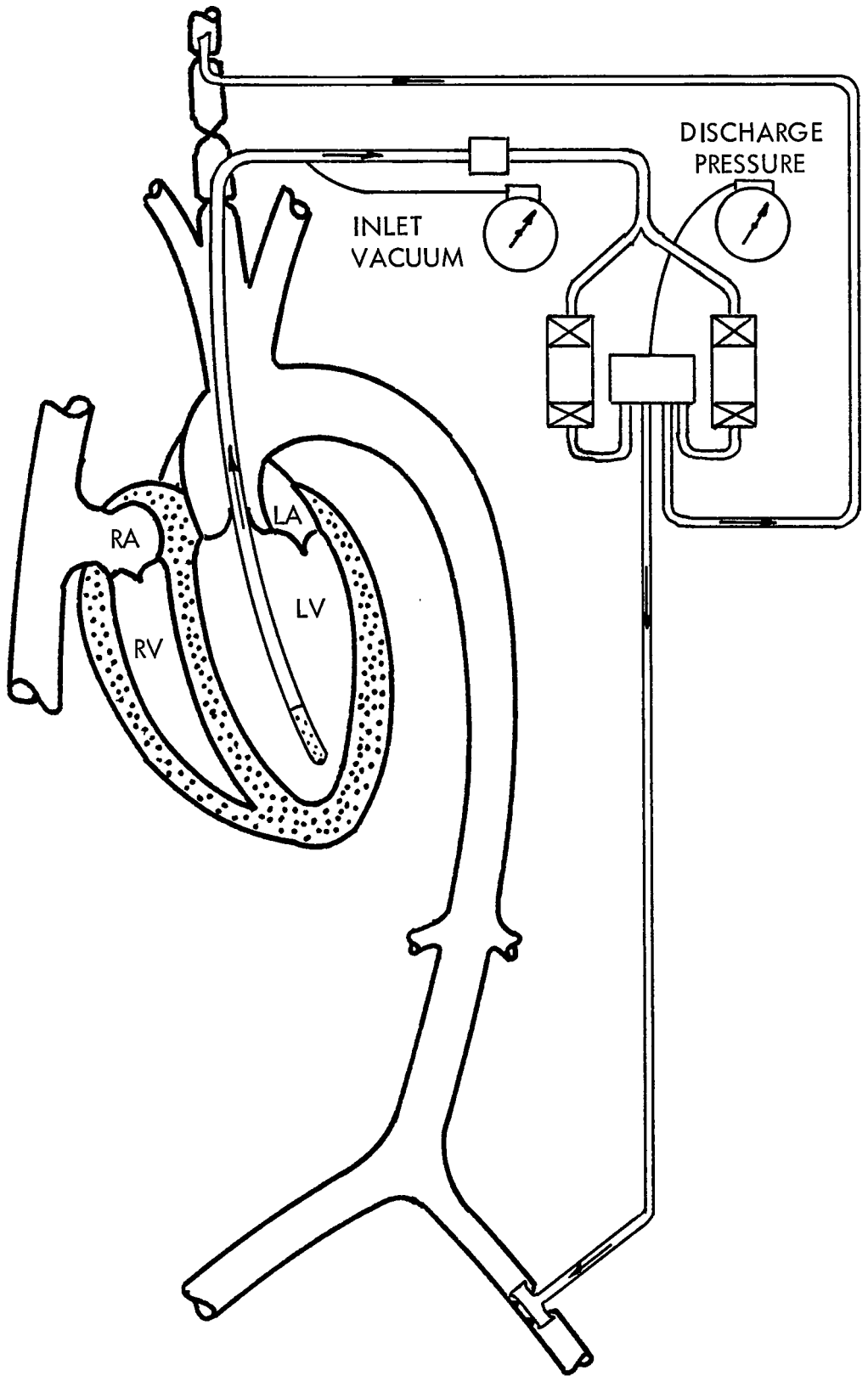


FIGURE 50

GEDDES-SCHUHMANN METHOD OF LEFT VENTRICULAR BYPASS

flow rates not be used in this catheter, because excess intracranial pressure may result.

During the developmental stages of the closed-chest left ventricular bypass method, pumping was provided by a roller pump (Sarns). Due to the controversy over the detrimental effects of non-pulsatile flow, later experiments have included the use of the aforementioned collapsible bag-type pump. Experimental work was carried out in mongrel dogs ranging in weight from 15 to 25 kilograms and involved graphic recording, via Physiograph, of many system variables as well as visual monitoring of others. A sample Physiograph record is shown in Figure 22.

The immediately apparent advantage is the minimum of surgery necessary for application of the method. The only incisions necessary are in the region of the neck for the right carotid artery and the groin area for the femoral artery; local anesthetic will permit gaining access to these vessels, thereby eliminating the danger of general anesthesia in the critically ill subject. This amount of surgery is matched only by the methods of counterpulsation and atrial septal puncture. Adequate flow rates are difficult to achieve in the counterpulsation method if only peripheral blood vessels are used, and correct adjustment of the system is difficult. The atrial septal puncture method has some attractive features but its use creates physical damage to the interatrial septum.

Another major advantage of the method of closed-chest left ventricular bypass is not quite so apparent. It is well known that the work output of the

heart can be measured by the oxygen consumed by the cardiac muscle. Thus, a "resting heart" consumes less oxygen than a "working heart." Rodbard et al. [80] have shown that the myocardial oxygen consumption is primarily determined by the tension in the heart muscle. The important point is this: If the ventricle is not continuously emptied by some means, blood will collect there, by means of the Thebesian vessels, until the pressure within the ventricle is sufficient to open the aortic valve. Thus, even if the major part of the blood bypasses the ventricle, the left ventricle must do non-flow work on the bolus of blood entrapped therein. This is also elucidated by Shenk et al. [89]. There are only two methods known to date in which this problem is solved by continuous withdrawal of blood from the ventricle. One is the method herein described, while the other is the method of Filler et al. [31]. The latter of these requires thoracotomy and puncture of the apex of the left ventricle.

Based on these premises, the investigators at Baylor felt that they had sufficient reasons for development of the method. Even so, the method is not totally without its disadvantages. Probably the major disadvantage appears at the inlet line to the pump. The ventricular catheter used, which is the pump inlet line, is restricted in size according to the diameter of the right carotid artery. In our experimental work on dogs weighing approximately 20 kilograms, a piece of Teflon tubing 10 inches long with an outside diameter of 4.3 millimeters (mm) was as large as could be used. This tubing has an inside diameter of 3 mm and seriously restricts the volume of fluid which can be pumped and

produces a high flow velocity during total circulatory support.

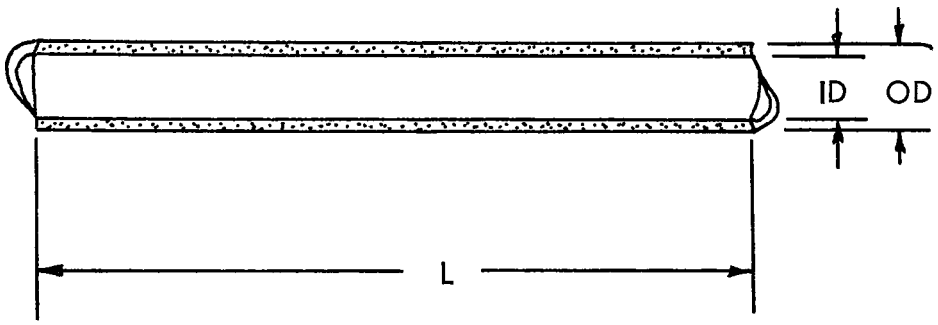
Another feature of the method which requires some care (and skill) in implementing is the installation of the ventricular catheter. Special care must be exercised during entrance of the catheter into the ventricle so that the aortic valve leaflets are not damaged. At one time the question of competency of the aortic valve with the catheter in place was raised. This is of utmost importance since operation of the system requires a competent valve. Experimental evidence, however, showed that the animal's cardiac output (as calculated by the saline injection method of Smith et al. [91, 92]), immediately before and after removal of the catheter, was unchanged. Patency of the catheter tip within the ventricle is ensured by means of a specially designed multiholed tip. If the catheter is properly placed, no problems of occlusion of the tip by the endocardium, papillary muscle or cordae tendinae have been found.

It is thought that the advantages of this system greatly outweigh the only real disadvantage (small inlet catheter). The installation of the ventricular catheter, after some practice, is routine and seldom, if ever, causes valve damage. The method has been applied in many experimental animals and shows great promise of clinical use in the treatment of heart failure.

APPENDIX B

PRESSURE-FLOW CALCULATIONS, INLET CATHETER

Due to the restricted inlet effect imposed on the pump by the small diameter inlet catheter, it is now desirable to determine the pressure-flow relations for this catheter. A sketch, with dimensions, of this catheter is shown below in Figure 51. The tip of this catheter is not shown nor is the pressure drop due to this tip considered. The open area of the tip is many times the flow area of the catheter, and experimental evidence has shown no additional flow restriction.



Length (L) = 26 cm. Outside Diameter (OD) = 0.43 cm.

Inside Diameter (ID) = 0.30 cm. Material - Extruded Teflon

$$\text{Flow Area (A)} = \frac{\pi (ID)^2}{4} = 0.0706 \text{ cm}^2$$

FIGURE 51

INLET CATHETER

The head loss, or pressure drop, due to fluid friction and fluid-wall interactions can be found in most texts on elementary fluid mechanics (see, for example, Vennard [94]). The equation for head loss in circular cross section tubes is given by

$$h_L = f \frac{L}{D} \frac{V^2}{2g} \qquad \text{(B-1)}$$

where h_L = head loss (cm of fluid),

f = friction factor (dimensionless),

L = tube length (cm),

D = tube inside diameter (cm),

V = average flow velocity (cm/sec),

g = acceleration due to gravity (cm/sec^2).

This expression gives the head loss in terms of cm of blood; a more desirable dimension would be mm of mercury. A conversion factor is now needed to get from cm of blood to mm Hg. Taking 1.055 gm/cm^3 as the density of blood (ρ_B) and 13.6 gm/cm^3 as the density of mercury (ρ_{Hg}), it is seen that one cm of blood is equivalent to 0.776 mm Hg.

$$1 \text{ cm blood} = \frac{1.055}{13.6} = 0.0776 \text{ cm Hg} = 0.776 \text{ mm Hg} \qquad \text{(B-2)}$$

The head loss in terms of mm Hg now becomes

$$h_L = 0.776 \, f \frac{L}{D} \frac{V}{2g} \qquad \text{(B-3)}$$

According to Vennard [94, p. 192], the friction factor (f) for a smooth

pipe with Reynolds Numbers (Re) between 3,000 and 10,000 can be expressed by

$$f = \frac{0.316}{(Re)^{0.25}} \qquad (B-4)$$

The Reynolds Number is a dimensionless parameter defined for a tube by

$$Re = \frac{VD\rho}{\mu} \qquad (B-5)$$

where V = average flow velocity (cm/sec),

D = inside diameter (cm),

ρ = fluid density (gm/cm^3),

μ = fluid viscosity (gm/cm sec).

According to Whittaker and Winton [101], the viscosity of blood is dependent on hematocrit (Hct.) and varies from 3 times that of water at a hematocrit of 30 to 5 times that of water at a hematocrit of 50. These hematocrit values span the normal physiological values. At 100°F the viscosity of water is 0.00679 poise (gm/cm-sec), yielding values for blood viscosity of 0.0364 and 0.0218 poise at hematocrits of 50 and 30, respectively. We can now write the values of Re in terms of the velocity using the aforementioned values for ρ, μ and D.

$$\left.\begin{aligned} Re\ (Hct. = 30) &= \frac{VD\rho}{\mu} = \frac{(V)(0.30)(1.055)}{0.0218} = 14.5V \\ Re\ (Hct. = 50) &= \frac{VD\rho}{\mu} = \frac{(V)(0.30)(1.055)}{0.0364} = 8.70V \end{aligned}\right\} \qquad (B-6)$$

It is also known that the flow rate (Q) is the product of average velocity (V) and cross sectional area (A), that is

$$Q \left[\frac{cm^3}{sec}\right] = V \left[\frac{cm}{sec}\right] A \left[cm^2\right] \quad OR \quad V = \frac{Q}{A} \qquad (B-7)$$

For our catheter the average velocity becomes the following, using flow in liters per minute,

$$V = \frac{Q}{A} = \frac{1000}{60} \frac{Q}{0.0706} = 236\,Q \left[\frac{l}{min}\right] \qquad (B-8)$$

The Reynolds Number now becomes

$$\left.\begin{aligned} Re\ (Hct. = 30) &= 14.5V = 14.5\ (236\,Q) = 3420\,Q \\ Re\ (Hct. = 50) &= 8.7V = 8.7\ (236\,Q) = 2060\,Q \end{aligned}\right\} \qquad (B-9)$$

The head loss as a function of flow rate can be calculated using equations B-9 and B-3. A summary of these calculations is shown in Table VIII and graphically in Figure 52.

According to Coulter and Pappenheimer [18], turbulence in flowing blood occurs near a Reynolds Number of 2000. Thus, at flow rates above 0.5 l/min, flow within the catheter is probably turbulent and the above analysis holds. However, at lower flow rates, the flow would probably be laminar and the above analysis would be questionable. The head loss for laminar flow is given by

$$h_L = [mm\ Hg] = 0.776 \frac{32\mu LV}{\rho g D^2} = 0.776 \frac{8\pi\mu L}{\rho g A^2} Q \qquad (B-10)$$

where all parameters are as previously defined. Using the two viscosity values

TABLE VIII

PRESSURE-FLOW RELATIONS, INLET CATHETER
TURBULENT FLOW

Blood Viscosity (poise)	Flow Rate (l/m)	Reynolds Number (Re)	Friction Factor (f)	Average Velocity (cm/sec)	Head Loss (mm Hg)
0.0218	0.0	0	------	0	0
	0.5	1710	0.0500	118	23.9
	1.0	3420	0.0414	236	79.0
	1.5	5130	0.0371	354	160
	2.0	6840	0.0348	476	267
	2.5	8550	0.0328	590	392
	3.0	10260	0.0313	708	540
0.0364	0.0	0	------	0	0
	0.5	1028	0.0560	118	26.8
	1.0	2060	0.0470	236	89.8
	1.5	3080	0.0425	354	183
	2.0	4110	0.0395	476	299
	2.5	5140	0.0374	590	448
	3.0	6170	0.0356	708	611

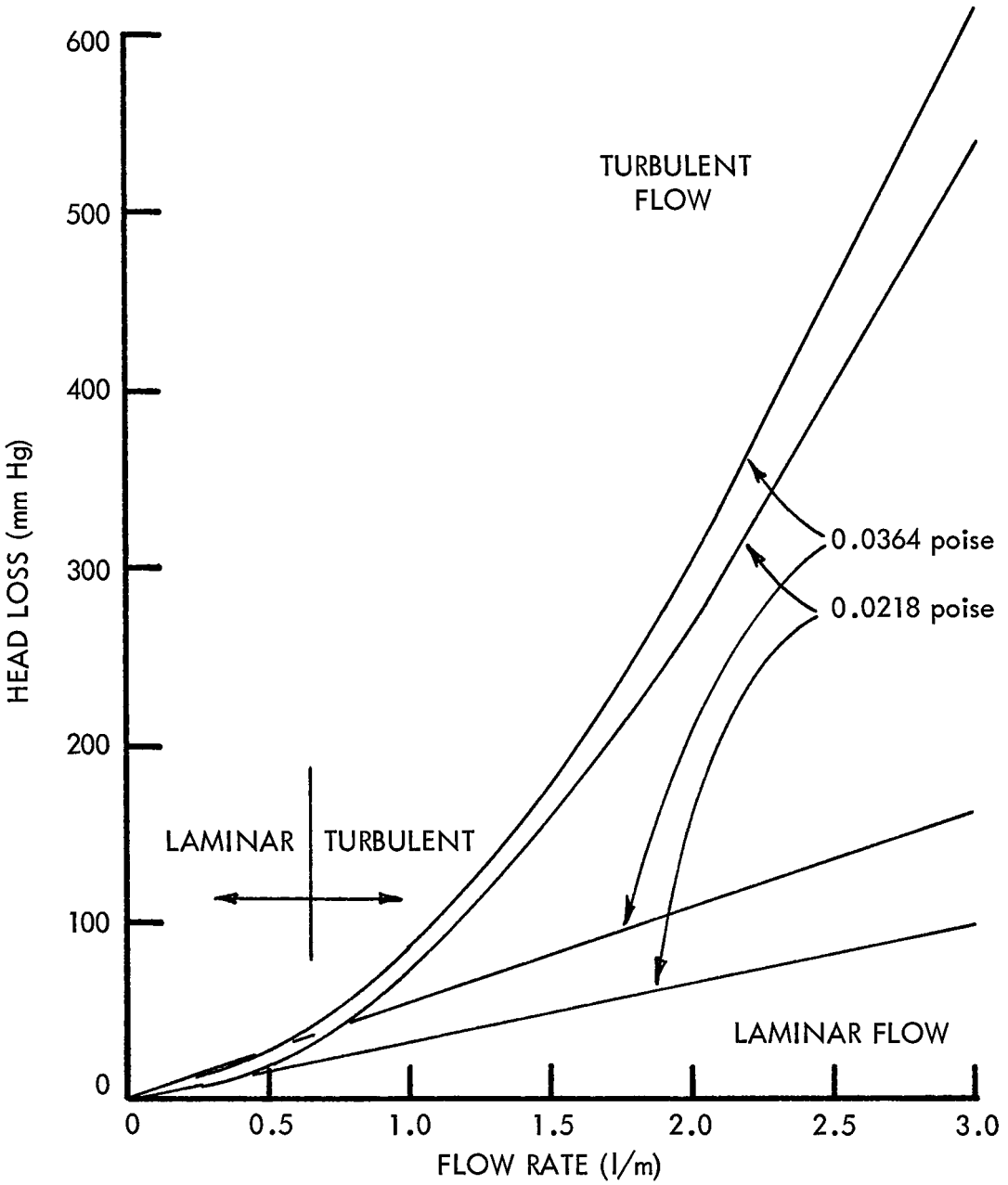


FIGURE 52

HEAD LOSS VERSUS FLOW RATE, INLET CATHETER

of 0.0218 and 0.0364 poise, this equation becomes

$$\left.\begin{aligned} h_L\ (\text{Hct.} = 30) &= 30.8Q\ [\text{l/min}] \\ h_L\ (\text{Hct.} = 50) &= 51.4Q\ [\text{l/min}] \end{aligned}\right\} \qquad \text{(B-11)}$$

These equations are also plotted on Figure 52 and show that at flows less than 0.5 l/min the head loss is nearly the same as that calculated from equation B-3. Two facts should be kept in mind when reviewing this information. First, the head loss is a pressure drop along the catheter and remains the same whether the driving force is a pressure at one end of the tube or a vacuum at the other end. Secondly, the analysis assumes steady-state, non-pulsatile flow whereas actual flow conditions are somewhat pulsatile. If pressures and flows used are mean values, however, the results are probably adequate to explain the phenomenon.