FLOW VISUALIZATION STUDIES OF A LEFT VENTRICULAR BYPASS PUMP

by

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ABSTRACT

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A flow visualization method was implemented in the study of the fluid flow characteristics of a swirl type left ventricular bypass pump. A systematic combination of pump spatial inlet orientations, pump valves, and pump body geometry were examined in order to determine pump flow characteristics. Both still and high speed motion photography were made of operating, transparent pumps in a mock circulatory loop with a blood analogue fluid. Regions of pump stasis, low velocity flows, and eddy formation were identified. In redesigning the bypass pump, several changes were implemented that stemmed from these flow visualization studies. The final pump design incorporated an angled, tangential inlet duct with a tilting disc valve and a tilting disc valve in the outflow tract. A low profile, non-occlusive diaphragm was used with a nozzle shaped pump body to improve internal pump flow. These modifications should minimize areas of possible clot formation and fibrin deposition in the pump.
ACKNOWLEDGEMENTS

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CHAPTER I

Introduction

Development of a clinically useful left ventricular bypass pump (LVBP) has been the common goal of researchers from Baylor College of Medicine and Rice University since 1967. This particular type of parallel left heart assist device is used to relieve the workload of the failing left ventricle. This is accomplished by shunting blood from the left atrium to either the descending thoracic aorta or the axillary artery (see Figure 1.1). This pump has been used clinically with limited success on patients who, after undergoing an open chest surgical procedure, could not be taken off the heart-lung machine.\(^1\)

Most of the problems associated with bypass pumps have been (1) fibrin buildup within the pumping chamber which eventually interferes with pump function, (2) high rates of hemolysis, and (3) thromboembolic complications. Cost of clinical use is an additional consideration.

In seeking an improved pump design, researchers have investigated various biomaterials, prosthetic valves, pump geometries and modes of pump operational control in an attempt to solve these problems.
PNEUMATIC AIR DRIVE UNIT

FIGURE 1.1

FIGURE 1.2
Flow visualization techniques have been useful in the past, not only in studying dynamic characteristics of prosthetic heart valves, but also in studying a limited number of blood pumping systems. These techniques have been useful in locating specific regions of severe turbulence, flow stasis, poor "washing" and eddy formation within a test section. Fibrin deposition and thrombus development have been associated with these types of fluid flow.

Vadot, Liotta, DeBakey, Ross and Wieting, have reported a "whirl" or "swirl" type bypass pump which appeared to reduce regional pump stasis due to the vortical nature of its internal flow. This type of bypass pump shown in Figure 1.2, used a dacron velour lined, silastic diaphragm which displaced approximately 50 - 60 cc of blood with alternate applications of positive and negative air driving pressures. Two one-way check valves were used to direct flow through this LVBP.

In subsequent experimental procedures using calves as the pump recipients, clot and fibrin deposition occurred within these devices which appeared to be still related to flow stagnation or to low velocity flow fields. Some examples of undesirable clot formation in the pumping chamber may be seen in Figures 1.3, 1.4, and 1.5.

In Figure 1.3, fibrin and thrombus deposition lines the pumping chamber and diaphragm of the pump and almost occludes the outflow tract. Massive clot buildup is evident
in the apical region of the bypass pump's polyurethane diaphragm shown in Figure 1.4. A close-up view of a Lillihei-Kaster inlet valve in Figure 1.5 shows fibrin ingrowth into the pivot points of the valve causing valvular dysfunction. 

In approaching this problem of clot formation and fibrin deposition within the bypass pumps, a flow visualization method was developed to evaluate and to redesign the modified swirl type left ventricular bypass pump. This experimental tool was implemented in the study of various pump geometries and pump valve combinations in order to develop a pump which minimized pump stasis and turbulent flow regions, and thereby reduced the number of possible sites for thrombus formation.
CHAPTER II

Initial Statement of Problem

Presented below is a method of evaluating the fluid flow characteristics of a swirl type left ventricular bypass pump. In order to design a pump with the most desirable flow characteristics, several combinations of experimental variables were considered. These variables were (1) pump inlet spatial orientation, (2) pump inlet and outlet valve type, and (3) pump body geometry.

The pump inlet spatial orientations that were considered are shown in Figures 2.1 and 2.2. Figure 2.1 shows three possible rotations of the inlet connector in a vertical plane passed through the pump. Figure 2.2 illustrates three possible rotations of the inlet connector in a horizontal plane passed through the pump.

The valves used in this study were selected from commercially available prosthetic heart valves, with the exception of one in-house designed pivoting disc valve. These valves, as may be seen in Figure 2.3, included the Lilliehei-Kaster tilting disc, the DeBakey aortic ball valve, the Kay-Shiley central occluder disc valve, and a pivoting disc valve.

Two pump body geometries were considered in this
study. The conical shaped pump body (a 30° cone), shown in Figure 2.4(a), was used extensively in the testing of the valve-inlet connector combinations. The nozzle-type body shown in Figure 2.4(b) was considered only in later experimental development of the bypass pump.

As is readily apparent, testing each type of valve in each of the pump inlet and outlet configurations would have been a monumental undertaking. The reduction of the number of possible test configurations was initially accomplished by these four experimental simplifications.

(1) Only horizontal and angled inlet connector configurations were to be considered (Figures 2.1(a) and (b)). The angle of inlet connector, as shown in Figure 2.1(c), was not considered because of the obvious flow stagnation fields that could develop in the pump's diaphragm region and because of inflow interference with outlet valve closure.

(2) Initially, only one type of pump body geometry, the conical shape, was to be considered. This geometry was much easier to machine and its conical shape was identical to the modified swirl LVBP that had been used in previous animal experiments. This would allow for more accurate correlations between flow studies and \textit{in vivo} studies to be made.

(3) A simple two-dimensional flow visualization model was developed to study the interrelated effects of pump inlet connector angle in the horizontal plane of rotation and the
effects of inlet valves on pump fluid characteristics.

(4) Valves in the inlet position were to be mounted as close as possible to the pump body to insure good "washing."

Initial Flow Visualization Experiments on the Pump Inlet

Three cross sections of different inlet angles in the horizontal plane were machined from 1/2" transparent Plexiglas sheets. As shown in Figures 2.5 and 2.6, these interchangeable cross sections were bolted onto a black Plexiglas base which contained an inlet reservoir(a) and an outlet port(b) modified by a vortex spoiler. This vortex spoiler was used to eliminate vortical flow caused by test chamber geometry. The spoiler consisted of a flat plate with numerous small drainage holes.

Pump valves, as they would appear in two-dimensional cross section, i.e., a ball valve would appear as a cylinder, and a disc would appear as a finite flat plate, were machined out of brass. An adjustable brass arm (c) with an adjustable valve attachment was connected to the side of the apparatus so that any combination of inlet cross section valve type could be simulated.

Tap water supplied the inlet reservoir with an almost constant flow and pressure head. As water filled the reservoir, liquid would flow horizontally through the inlet cross
FIGURE 2.6

FIGURE 2.7

Two Dimensional Flow Study Model
section, into the simulated pump, and finally out of the apparatus through a second control valve. The length of the inlet connector was long enough so as to visually confirm a uniform velocity profile proximal to the idealized two-dimensional valve. The dimensions of the inlet connector and pump cross section were made twice as large as the experimentally used swirl LVBP to facilitate flow field determinations.

Flows in this idealized two dimensional LVBP model were made visible by mixing spherical, reflective Amberlite resin beads, (0.007 - 0.0165 inches diameter) in the inlet reservoir. These beads have an average density of 1.1 gm/cc, but many were approximately neutrally buoyant and were carried along with the fluid. A 250 watt flood lamp and shade were used to illuminate the flow field. This was accomplished by placing the flood lamp perpendicular to the flow field as shown in Figure 2.7.

A Honeywell Pentax SP 500 (1:2/55 lens) camera with a 2X converter was mounted on a tripod directly above the experimental flow apparatus. Using Kodak Tri-X film (ASA 400), black and white photographs were made at shutter speeds ranging from 1/8 to 1/30 second. Approximately the same flow rate was established for each combination of inlet cross section and valve. General pump steady flow characterizations could be determined from these photographs. No exact determination of the flow rate was obtained; neither were
precise velocity field determinations nor a dynamic similarity correlation, such as a Reynolds number, made between the actual LVBP and the flow visualization model. The objective of these experiments was only to determine the steady state pump flow fields. Results of these experiments are presented in the next section of this paper.

Results of the Two Dimensional Flow Studies

Results of the two dimensional flow studies are presented in the following manner. In all, three different types of inlet connectors were considered: tangential, off-center, and central inflow. Photographs of each inlet orientation were grouped together. Within each group, the effects of the three different valves were demonstrated. In describing these pump flows, higher velocity fields within the pump may be noted because the streaks traced by the particles were longer. Regions of pump stasis and low velocity flows appeared as shorter streaks or as points. Streamlines were coincident with the tangent of the velocity vectors of the fluid particles in steady flows. A streakline is defined as a line formed by fluid elements which pass through a given location in a flow field. Streaklines were useful in pulsatile flow conditions. Again, in steady flow, the streaklines coincided with the streamlines.

Discussions of the flow fields within the pump follow; along with the pros and cons of each valve inlet arrangement.
A summary statement with preliminary conclusions appears at the end of this section.

Figure 2.8(a) shows the effects of a tangential inflow with a tilting disc valve. As is readily apparent, a vortical flow within the pump was established. Good "washing" of the sides of the pump is demonstrated. A small region of stagnation occurred on the tip of the disc valve because of its blunt body construction. Normally, prosthetic disc valves of this variety are more streamlined and tend to eliminate much of this stagnation. Stagnation was also noted in the center of the pump. Because this flow was vortical in nature, one would expect a small region of stasis as in the eye of a hurricane. Velocities directly at the wall were zero as is true in any viscous fluid flow.

Next considered are the effects of an occluder disc valve in a tangential inflow. As may be seen in Figure 2.8(b), doublet formation is apparent. Stagnation occurred at the centers of the doublet pair. High velocity flows existed at valve openings while a large region of stasis occurred behind the disc valve. A smaller low velocity area was found near the pump wall at about four o'clock.

In Figure 2.8(c), a ball valve in the tangential inflow position is shown. The regions of stasis were similar to those which occurred with the occluder disc valve. Areas of stasis or stagnation were present proximal and distal to the valve and again at four o'clock.
FIGURE 2.8 (a)

FIGURE 2.8 (b)
Figure 2.9 shows the effects of valves on the pump flow field established by an off-center inlet connector. Again, with a pivoting disc valve (Figure 2.9(a)), a vortical type of inflow was established within the pump cross section. However, because of the off-center geometry of the inlet connector, a large region of stasis occurred in the upper portion of the cross section. Flow separation again was seen on the trailing edge of the blunt valve. Flow elsewhere in the pump adequately "washed" the walls.

Stagnation regions immediately behind the occluder disc-valve are demonstrated in Figure 2.9(b). Another region of low velocity flow was present about seven o'clock on the pump wall. Again doublet flow formation was noted.

Figure 2.9(c) shows some interesting effects of a ball valve on the off-center inlet geometry. High velocity jetting was seen around the ball valve. This jetting, in combination with this geometry produced a region of stasis along the wall between nine and twelve o'clock. Flow separation off the ball valve occurred at about 110°. Another small region of flow stasis was present about three o'clock, just below the ball valve.

The next pump geometry considered (Figure 2.10), aligns the inflow directly toward the center of the pump cross section. Figure 2.10(a) shows the massive stagnation region produced using a tilting disc valve. A swirl flow field was established, but was not of sufficient strength
to "wash" the pump surfaces. Flow separation occurred at the intersection of the inlet connector and the pump body.

An occluder disc valve shown in Figure 2.10(b) established a doublet flow field with stasis and stagnation regions at one and seven o'clock. In Figure 2.10(c), the effect of a ball valve may be seen. Prominent regions of stasis were present between eleven and one o'clock and three and five o'clock near the pump's sides. Vortices were also seen separating from the distal side of the ball valve. Vortical flow was the prominent flow feature in the rest of the pump cross section.

Summary and Conclusions

The geometry and valve combination that produced the least amount of pump stagnation and stasis employed a tangential inflow incorporating a tilting disc valve. In the remaining eight test combinations, either the pump inlet orientation or valve type established undesirable flow characteristics in the pump. These regions of stasis were possibly the in vivo source of thrombo-embolic complications.

Figures 2.11 and 2.12, are side by side comparisons of in vitro flow visualization experiments and in vivo LVBP experiments in calves. Pump inflow geometry is very important because 60% to 80% of the pump's cycle time is used in filling the pump across a low pressure gradient.
CHAPTER III

LVBP Pulsatile Flow Studies

Described in this section is a method by which pulsatile flow studies were made on left ventricular bypass pumps. These studies were performed to obtain accurate information on the nature of the flow within an operating bypass pump.

The experimental apparatus is shown in a photograph in Figure 3.1(a) and is illustrated in Figure 3.1(b). The major components of this experimental set-up included (1) the transparent bypass pump models, (2) the pneumatically powered drive unit, (3) the mock circulatory loop and (4) the blood analogue fluid. Each of these components will be described separately.

The bypass pump models were machined from blocks of Plexiglas to facilitate internal viewing of the pump flow fields. Both the inside and the outside faces of the machined blocks were polished with pumice stone to improve transparency. Since the refractive properties of the blood analogue fluid and the Plexiglas were similar, little distortion of the internal flow was noted.

The original flow visualization pump models were designed to have similar dimensional characteristics of the
experimentally used swirl bypass pumps. This was done to compare *in vitro* tests with *in vivo* data.

Three geometrically different pumps were considered. These may be seen in Figures 3.2, 3.3 and 3.4. In Figure 3.2, the inlet duct is either perpendicular to the pump's conical body or perpendicular to the pump's vertical axis. Figure 3.3, shows the inlet duct tangential to the cone of the pump while in Figure 3.4, the inlet connector directs flow toward the center of the pumping chamber. The inlet and outlet connectors in these models were designed to accept a variety of valve types. These three pump geometries were chosen because (1) the angle of the inlet duct in the vertical plane was yet to be determined and (2) verification of the first two-dimensional flow visualization study was necessary.

Three of the four valves used in this study were commercially available prosthetic heart valves. Each was furnished with a cloth sewing ring. Removal of this sewing ring and its replacement with a Plexiglas clip-ring was necessary to mount these valves in a stable position in the bypass pump. Figures 3.5(a) and (b) show typical flow visualization model.

The diaphragm used in these pumps was usually made from polyurethane. This was made by repeatedly dipping an epolene wax diaphragm mold into liquid polyurethane until the proper thickness was achieved. This material was
selected because of its relative transparency and because of its use in experimental pumps. Silastic diaphragms were also implemented in this study, but no attempt was made to distinguish flow field effects caused by the diaphragms' property differences.

The pneumatically powered drive unit\(^{(5)}\) used in this study was designed and built at Rice University for use in its artificial heart program. It provided the following pneumatic characteristics (1) pulse rates from 1-99 pulses per minute (2) driving air pressure from 0-250 mmHg and vacuum from 0-150 mmHg and (3) variable systolic durations from 1-990 milliseconds.

The mock circulatory loop was of simple design. It included a pressurized outlet reservoir, rotometer, inlet reservoir, inlet reservoir float valve and connecting tubing (see Figure 3.1(b)). The systemic pump impedance that the bypass pump experienced was grossly simulated by the pressurized outlet reservoir. Both the inlet and outlet reservoirs were constructed from eight-liter plastic containers. Variable static systemic pressures were regulated by a non-relieving air pressure regulator (modified Fairchild, Model 10, 0-10 psig).

In briefly tracing flow through the loop, outflow from the outlet reservoir flowed through the rotometer (Dwyer, Type RMC-143) and into the inlet reservoir through a styrofoam float valve. This float valve provided a
constant inlet level control. The tested bypass pump was connected between the inlet and the outlet reservoirs with tygon tubing.

The blood analogue fluid was an aqueous-glycerin solution which contained 39% glycerin (USP) by volume and 61% tap water. This blood analogue fluid provided an excellent flow visualization medium as reported by Wieting and Pierce. The viscosity of this solution was approximately 3.5 centipoise at room temperature. Its specific gravity was 1.1. This compares to the average viscosity of whole human blood of about 3.33 centipoise and specific gravity of 1.06. This aqueous-glycerin solution behaves as a Newtonian fluid (i.e., the viscosity does not vary with shear rate) as does whole human blood in flow in larger arterial vessels.

**Flow Visualization Equipment**

The method of flow visualization involved the use of (1) tracer particles suspended in blood analogue fluid, (2) a slit light source, (3) a 35 mm Honeywell Pentex SP500 camera and (4) a Hycam High Speed camera, Model 41-0004 (film speeds 10-11,000 frames per second).

Flow patterns were made visible by suspending spherical Amberlite(R) ion-exchange resin beads in the blood analogue fluid. This technique was successfully used by
These spherical beads were prepared in the following manner. A mixture of Amberlite\textsuperscript{(R)} was first screened to obtain a uniform size particle, then, because the Amberlite\textsuperscript{(R)} was slightly less dense than the analogue fluid (1.08 gm/cc to 1.1 gm/cc), the beads were soaked in mineral oil overnight to increase their density. They were then allowed to stand for one hour in a small container of aqueous-glycerin. Beads that continued to float were skimmed from the top, while the remaining particulate suspension was added to the mock circulatory loop. This suspension of particles provided good photographic results.

Flow patterns in almost any plane of the bypass pump were made visible by a narrow beam of light. A narrow beam of intense light was obtained using a Par light fixture (Model 500 WFL) adapted with (1) a polished parabolic mirror, (2) a 1000 watt quartz diode element and (3) two plano-convex cylindrical lenses. The intense beam of light so generated was passed through a variable slit arrangement. The light was powered by a dc power supply at 120 volts. This narrow slit of light was focused on the transparent pump housing and provided planar flow information.

Because the slit light technique allowed one to see only motion in the illuminated plane and no motion in the lighted plane perpendicular to the viewer (i.e., fluid particles moving toward the viewer), an additional flow
visualization technique was also used in evaluating flow fields. Air was injected into the inlet connector just proximal to the pump's inlet valve. These air bubbles were useful in determining aspects of the three-dimensional pump flow including the vortical flows associated with some of the pump-valve combinations.

Cross sectional views of the flow looking through the diaphragm were difficult to obtain because of its semi-opaque qualities. To obtain such two-dimensional flow fields, the diaphragm of the tested pump was removed. Another diaphragm type bypass pump of comparable stroke volume was connected serially with the tested LVBP. This arrangement, as seen in Figure 3.6, provided pseudo-pulsatile flow through the transparent flow model.

In making films of operating bypass pumps, filming speeds from 48 to 240 frames per second were used. The film chosen was 16 mm Kodak 4X Reversal, Black and White (ASA 320 Tungsten). Normal motion picture projector film speed is 24 frames per second. Higher film speeds allowed for a frame analysis of the flow in which visualization particles appeared as dots or short streaks. Reproduction of individual frames of slower film with longer streakline information was convenient in presenting still slides on the dynamic behavior of the pump. Films made at the higher film speed, when viewed with a 16 mm movie projector, presented continuous slow motion visualization of flow studies.
Transparent LVBP Flow Studies

In describing many of these LVBP movies in this thesis, one sometimes was limited to a written analysis of the pump flow. Other flow descriptions were accompanied by single frame reproductions or freehand art work. This mode of presentation was necessary because of the difficulty in presenting cine information in a bound volume.

Most of these films were made under similar air drive parameters, inlet-outlet pressure heads, and connecting tubing lengths. Average pumping parameters are given below:

<table>
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<th>Parameter</th>
<th>Value</th>
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<tr>
<td>Pump Rate</td>
<td>60 pulses per minute</td>
</tr>
<tr>
<td>Pump Systolic Duration</td>
<td>270 milliseconds</td>
</tr>
<tr>
<td>Pump Air Drive Pressure</td>
<td>120 mmHg</td>
</tr>
<tr>
<td>Pump Air Vacuum Pressure</td>
<td>0-5 mmHg</td>
</tr>
<tr>
<td>Pump Flow Rate</td>
<td>3.0-4.2 liters per minute</td>
</tr>
<tr>
<td>Pressure Gradient Across Inlet Valve (Diastolic)</td>
<td>15-20 mmHg</td>
</tr>
<tr>
<td>Pressure Gradient Across Outlet Valve (Systolic)</td>
<td>40 mmHg</td>
</tr>
</tbody>
</table>

In the first film to be described the following pump geometry was implemented. The inlet duct is tangential to the pump cone and perpendicular to the pump's vertical axis. A Lillihei-Kaster tilting
disc valve was used in the inlet position while a DeBakey ball valve was mounted in the outflow tract. A polyurethane diaphragm was also used. A slit light source parallel to the inlet duct was implemented.

In viewing the film, one observed no major regions of pump flow stasis except during filling. As seen in Figure 3.7(a), a ring vortex was formed in the apex of the diaphragm at the end of pump diastole. This vortex formation was probably due to the angle of inclination of the inlet duct, i.e. it was directed toward the pump wall and not toward the pump's diaphragm region. At the end of pump systole (Figure 3.7(b)), one also noted the entrapment of fluid between the diaphragm and the pump body in the region of diaphragm attachment. This collapse of the diaphragm on the pump walls not only created a massive ring-shaped region of stasis, but also caused undue trauma by crushing formed elements of the blood. In considering the complete pump cycle, reasonably good "washing" of the pump surfaces occurred.

Total displacement of the diaphragm was about 62 cc with a displacement distance of approximately 3 cm. This diaphragm underwent a number of foldings and unfoldings during the course of one pump cycle. These formed crevices were not only sites of additional blood entrapment, but also were regions of high diaphragm stress concentrations which may reduce its lifetime.
Another film was made of the same pump geometry and valve combination, however, the slit light was passed through a plane perpendicular to the inlet connector. The view of the camera, seen in Figure 3.8, is into the inlet valve. Again one noted generally good pump "washing." Complete flow circulation reversal occurred as the diaphragm moved to a fill position.

Another interesting observation was the erratic motion of the outlet ball valve. Even though flow out of the pump tended to push the valve into an open position, the ball bounced randomly from an almost seated position to a full open position during pump ejection. If the valve was immersed in the blood analogue fluid, the ball remained at the bottom of the cage in an open position. Also the annular space in the outlet flow tract had the same area as the pump outlet flow duct. Speculation of the reasons why the ball "bounced" will be discussed in the next film overview.

In the following film, the pump contained an inlet Lillihei-Kaster valve and an outlet DeBakey ball valve. The tangential inflow tract was perpendicular to the 30° cone. In addition to the suspended, reflective beads, air bubbles were introduced into the
inlet tube with a 10 cc syringe. The swirl or vortical nature of the pump flow was recognizable because of the concentration of air bubbles into the center of the pump. This type of vortical inflow was also described in a two-dimensional photograph, Figure 2.1(a). Other experimental evidence to support this theory is presented in another section. Six sequential photographs of pump systole are shown in Figure 3.9. As may be seen, the fluid's angular velocity increased as systole progressed. The core of air bubbles extended outward from the diaphragm and eventually attached itself to the outlet ball valve. During pump filling, the air bubbles remained in contact with the apex region of the diaphragm. Development of a strong vortex core seemed to correlate in time with the instability of the ball valve. In the center of the core of an inviscid, free vortex, fluid velocities approach infinity while core pressure approaches absolute zero.

If one considers the force balance shown in Figure 3.10, one may conclude that some pressure disturbance caused the ball to move toward a closed position during pump systole. Forces which acted in the positive Y direction were gravity, viscous drag, and the pressure gradient across the ball valve. Forces which acted in the negative Y direction were buoyancy and pressure disturbances.

Complete closure of the outlet valve did not occur during pump systole because as the ball began to seat, the
FIGURE 3.8

(a)  (b)  (c)

(d)  (e)  (f)

FIGURE 3.9
annular outflow area was reduced and the fluid velocity and viscous drag was increased. The ball valve therefore oscillated between states of equilibrium.

In any case, the ball appeared to be well "washed" by this swirl outflow except for stagnation regions located in the valve's cage and on the distal side of the valve, as shown in Figure 3.11. This swirl existed throughout pump diastole. Figure 3.12 shows the possible effects of this type of flow. Note the massive clot formation in this ball valve used in a LVBP calf experiment.

The distal side of the Lillihei-Kaster valve appeared to also be well "washed" by the swirling motion of the flow. However, the valve appeared to be easily caught in a partially open position by the wedging of the flow visualization spheres between the disc and valve seat. Hopefully clots introduced into the pump via the atrial cannula could be eliminated or are of insufficient quantity so as not to compromise this valve's function.

The effects of the position of the air drive line connectors on this inlet-valve combination were also considered. As may be seen in Figure 3.13, a second angle of inclination of the drive line was used to determine whether or not this change would effect the movement of the diaphragm and possibly the strength of the pump vortex.

The first fifty feet of a 100 foot role of movie film was taken with the air drive line attached to the angled
Flow

Bouyancy—Pressure
Disturbance

Gravity
Viscous Drag
Systolic Pressure Gradient

FIGURE 3.10

Flow

Ball Valve

FIGURE 3.11

FIGURE 3.12
inlet connector. Without changing the air drive pressure setting or other mock loop parameters, the air drive line was reattached in its normal position and the remaining fifty feet of film was exposed.

No difference was noted in diaphragm deflection or in pump flow during systole. However, during diastole, the diaphragm was deflected more toward the angled inlet air port than when the drive line was in its normal position. This movement accounted for no discernable change in the pump flow fields.

The next film to be considered showed the effects of an occluder disc valve mounted in an angled, off-center inlet duct and a DeBakey ball valve mounted in the outflow position. Several interesting fluid flow phenomena occurred as the result of using an occluder disc inlet valve.

During pump filling, high velocity flows occurred around the tip of the valve's disc as may be seen in Figure 3.14. Associated with this jet is a fluid phenomena as vena contracta, in which fluid velocities are elevated due to the minimization of the effective inlet cross sectional area. In Figure 3.15, one may also note fluid particles impinging and stagnating on the backside of the disc
FIGURE 3.13

Angled & Normal Air Drive Connector

FIGURE 3.14

FIGURE 3.15
and on its supporting struts. An area of stagnation also occurred on the pump wall opposite the inlet duct as a result of the doublet-like fluid flow seen in Figure 3.16(a). This photograph, taken with the diaphragm removed, as was described earlier, showed an area of fluid stagnation along the walls of the pump body. This corresponded well with the flow patterns seen in the idealized two-dimensional flow study shown in Figure 3.16(b). No swirl type inflow was developed during filling, but rather some doublet-like flow fluid field was established. Consequently, the vortical nature of the systolic period of the pump was not present as would have occurred with a swirl, reinforcing tangential inflow with the tilting disc valve. It was speculated that this is an explanation of why the ball valve was relatively stable during pump systole. While viewing these films one also noticed the occasional improper seating of the inlet disc valve (see Figure 3.17). This valve could be trapped in a half open position which would compromise the effectiveness of the pump.

Another film using the same valve-inlet connector combination was taken using a slit light passed through the pump perpendicular to the inlet valve. The observer's view was into the pump's inlet. One also noted the formation of a doublet pair during pump diastole. No vortical flow appeared in this pump view.
FIGURE 3.16(a)

FIGURE 3.16(b)
A pump incorporating a Lillihei-Kaster tilting disc valve in the tangential inlet position with a Kay Shiley occluder disc valve in the outlet position was the next pump-valve configuration to be considered. Vortical flow with good "washing" characteristics was established during both pump filling and ejection.

Air bubbles introduced proximally to the inlet valve again demonstrate the high velocity, low pressure core established during pump systole. As may be seen in Figures 3.18(a) and (b), stagnation of the vortex core on the proximal and distal surface of the disc was noted. Because of the strength of the vortex, the stagnation which appeared on the distal side of the valve was present during the majority of pump diastole.

High velocity flow fields near the valve were also noted. Higher pneumatic driving air pressure was necessary to achieve the same stroke volume with the same systolic duration used in previous experiments. This effect was caused by the decreased annular cross sectional area available for pump evacuation. All of these flow characteristics associated with this valve make it an undesirable choice in the outlet flow position.
The next pump valve arrangement to be considered used Lillihei-Kaster valves in both the inlet and outlet positions. The inlet duct was tangential to the pump body and also inclined toward the pump's diaphragm region. The outlet Lillihei-Kaster valve was mounted in two different ways. As may be seen in Figure 3.19, the Lillihei-Kaster was oriented with its seat perpendicular to the vertical pump axis. This resulted in a disc opening of approximately 80° with the horizontal. In Figure 3.20, the seat of the disc was set at approximately 10° with the horizontal. This resulted in the disc opening parallel to the outflow tract.

In viewing this film, the expected swirl type flow was established within the pump body. However, the amount of backflow one measured with the disc oriented parallel to the flow field (Figure 3.20) was much greater than the backflow associated with the valve arrangement in Figure 3.19. This information is shown on strip chart recording (Figure 3.21) from a square-wave electromagnetic flow probe distal to the pump. A Carolina Medical Electronics, EP 300, Series #174A was used. Even though a higher instantaneous pump flow rate was achieved with the disc parallel to the flow field, the amount of backflow through the valve was 10-15% than with the valve opening at 80°.

The in-house designed hinged disc valve, shown in a previous photograph (Figure 2.3), was used in the angled, tangential inlet position of the next filmed bypass pump.
A DeBakey ball valve was used in the outflow position. This type of hinged valve established a similar vortical inflow as does the tilting Lillihei-Kaster valve. The backside of the valve was well "washed" in the swirl which was present throughout the pump's cycle. As may be seen in this film (as was apparent in the other movies showing the inlet duct perpendicular to the pump cone), some interference with the inlet valve openings was noted due to the large axial displacement of the diaphragm. The "horns" or the supporting structure of the Lillihei-Kaster valve came in contact repeatedly with the diaphragm. This continual stabbing of the diaphragm could result in eventual puncture of the diaphragm with disastrous consequences. Outlet ball valve instability was again apparent.

If the ball valve was mounted into the inlet position as may be seen in Figure 3.22, it is readily apparent that the cage of the valve would severely interfere with the diaphragm's movement. This valve arrangement was not filmed. Results of the initial two-dimensional flow visualization studies and their confirmation in all pulsatile pump films lead to the conclusion that the ball valve would be a poor choice in the inlet position.

**General Observations of Pulsatile Pump Flow**

Several general comments may be made concerning information derived from these high speed films made of LVBP's.
First of all, any sharp corners associated with the inlet connector could cause flow separation and eddy formation that should be minimized so as to avoid small local accumulation of blood clots. This may be accomplished by the rounding of these corners.

Equally important were the effects of the motion of the diaphragm on the stability of both the inlet and outlet valves and on pump flow stagnation. The incoming fluid has some inertia and the diaphragm has some compliance. If the diaphragm is pulled back with too great a vacuum, the diaphragm will reach its end of travel and oscillate. This diaphragm rebounding will establish pressure disturbances within the pump's fluid which cause the pressure gradients across the valves to change with time. These changing gradients cause the valves to open and close unexpectedly. The same sort of valve instability may be seen in the outlet valve if the diaphragm is "overdriven" and allowed to oscillate at its end travel position.

One film record was made in which the driving pressures and pump systolic duration were so adjusted as to allow for continuous movement of the diaphragm. This resulted in minimizing end of travel accelerations and, therefore, diaphragm oscillations and valve instability.

A device developed by N. A. Normann of Baylor College of Medicine(12) electronically monitors the position of the
pump's diaphragm. With judicious selection of air driving parameters and system delay times, it is capable of providing a continuously moving diaphragm.

**Ball Valve Stability Experiments in Steady Flow**

As reported earlier in this section, the interaction between the ball valve and the vortical pump flow induced valve instability which could not be readily explained. A simple experiment was devised in order to correlate the low pressure, high velocity vortical flow with the valve instability.

A transparent bypass pump with a tangential inflow with a Lillihei-Kaster valve in the inlet duct and a DeBakey ball valve in the outlet position was mounted serially with a tap water faucet and a rotometer. Air bubbles were introduced into the system to aid in flow visualization. As may be seen in Figure 3.23, the air bubbles were concentrated in a small diameter core at the center of the pump. Steady flow rates of 0.5, 1.0, and 2.5 gallons per minute were established through the pump. Note in Figures 3.23(a), (b), and (c), as the flow rate increased, the pump vortex core grew smaller and the ball was lifted toward the valve's seat and remained in a stable position. The ball rested at the bottom of the cage in a no-flow condition. This simple test
FIGURE 3.23(a)

FIGURE 3.23(b)
FIGURE 3.23(c)

FIGURE 3.24
supported the idea that ball instability is related to the low pressure force in the center of this free vortex. Presented below are calculations which predict this experimental observation.

(1) The inlet angular pump velocity was determined as follows:

\[ \dot{q} = V_a A \]  

where \( \dot{q} \) = volumetric flow rate, liters per minute  
\( A \) = valve inlet cross section, cm\(^2\)  
\( V_a \) = inlet angular velocity, cm/sec

Solving for \( V_a \) yielded

\[ V_a = \frac{\dot{q}}{A} \]

Substituting experimental values of \( \dot{q} \) and \( A \) yielded

\[ V_a = \frac{0.704 \ \ell/\sec}{2.48 \ \text{cm}^2} \]

\[ V_a = 28.4 \ \text{cm/sec} = 0.932 \ \text{ft/sec} \]

This angular inlet velocity occurred at a pump radius, \( r_p \), of 1.08 inches as shown below:
The pressure distribution within an inviscid, free, unbounded vortex may be determined from the following equation:

\[ p_2 - p_1 = \left( \frac{1}{r_1^2} - \frac{1}{r_2^2} \right) \frac{\rho K^2}{2g_c} \]

where

- \( p_2 \) and \( p_1 \) = pressures with the vortex, psi
- \( r_2 \) and \( r_1 \) = radial positions within the vortex, inches
- \( \rho \) = density of water, lb/ft\(^3\)
- \( g_c \) = gravitational constant
- \( K \) = strength of the vortex, \( K = V_a \cdot r_p \)

Equation (2) was used to calculate the pressure distribution at the pump's exit where the pump's radius, \( r_2 = 0.35 \) inches. Tabulated below were the pressures obtained at various pump exit radii.
<table>
<thead>
<tr>
<th>Radial Position in the Vortex, inches</th>
<th>Differential Pressure Between Wall, r₂, and Some Radial Position, r₁, psi</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.050</td>
<td>2.629</td>
</tr>
<tr>
<td>0.075</td>
<td>1.200</td>
</tr>
<tr>
<td>0.100</td>
<td>0.618</td>
</tr>
<tr>
<td>0.200</td>
<td>0.113</td>
</tr>
<tr>
<td>0.350 (r₂)</td>
<td>0.000</td>
</tr>
</tbody>
</table>

(3) The outflow chamber, containing the ball valve, increased in radius from 0.35 inches to 0.55 inches as shown below:

Calculated next is the pressure distribution on the distal side of the ball. It was assumed that the wake behind the ball had a projected radius of 0.30 inches as seen below:
In determining the distal pressure distribution, the strength of the vortex, $K$, was assumed constant as was the pressure in the ball's wake. From equation (2), the differential pressure was calculated at $r_2 = 0.55$ inches. The table below presents these results.

<table>
<thead>
<tr>
<th>Radial Position in the Vortex, inches</th>
<th>Differential Pressure Between Wall, $r_2$, and Some Radial Position, psi</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.10</td>
<td>0.0525</td>
</tr>
<tr>
<td>0.20 [ in the wake</td>
<td>0.0525</td>
</tr>
<tr>
<td>0.30</td>
<td>0.0525</td>
</tr>
<tr>
<td>0.35</td>
<td>0.0326</td>
</tr>
<tr>
<td>0.45</td>
<td>0.0109</td>
</tr>
<tr>
<td>0.55 ($r_2$)</td>
<td>0.000</td>
</tr>
</tbody>
</table>

The graph below presents the proximal and distal pressure distributions about the ball valve.
The average pressure proximal to the ball was found by integrating this equation:

\[
P_{\text{AVE}} = \int_{0}^{r_2} p \, r \, dr
\]  

where:

- \( p \) = pressure
- \( r \) = radius
\[ P_{\text{AVE}} = \text{average pressure} \]

let \( p(r_2) = 14.7 \text{ psia} \)

The average proximal pressure was found to be 13.1 psia, while the average distal pressure was found to be 14.6 psia. The pressure difference across the sphere was

\[ \Sigma P = 14.6 - 13.1 \]

\[ \Sigma P = 1.5 \text{ psi} \]

This differential pressure acting on the projected area of ball \((r = 0.4 \text{ inches})\) would result in an upward lift of 0.75 lbf.

(5) Next was calculated the drag force on the ball using drag coefficients obtained from a sphere in parallel, unbounded flow. The exit velocity past the outlet valve was determined using equation (1).

\[ V = 2.33 \text{ ft/sec} \]

The Reynolds number was then calculated

\[ \text{Re} = \frac{V \cdot D}{v} \quad (4) \]

where:

\( V = \text{velocity past the sphere, ft/sec} \)

\( D = \text{diameter of ball, ft} \)

\( v = \text{kinematic viscosity of the fluid, ft}^2/\text{sec} \)
Re = 25900

The drag coefficient, \( C_D \), was found to be 0.45. Equation (5) was used to calculate the drag force on the ball

\[
\text{DRAG} = C_D \left( \frac{1}{2} \rho V^2 A \right) / g_c
\]  

(5)

where

\[ A = \text{projected area of the sphere, ft}^2 \]

Substituting in known information yields

\[
= (0.5)(0.45)(62.4)(2.3)^2(0.003491) / 32.17
\]

\[ = 0.0081 \text{ lb}_f \]

The pressure force across the ball is 93 times greater than the drag force. The net upward lift on the ball is 0.742 \( \text{lb}_f \).

In a similar test shown in Figure 3.24, a Kay-Shiley occluder disc valve was placed in the inflow duct. With flow rates as high as three gallons per minute, no outlet ball valve instability or lifting was noted. As reported earlier, this correlated with observations made in the pulsatile flow films.

To insure that the construction of the outflow tract played no significant part in the development of the ball valve instability, a conical shaped outlet tract was
constructed and tested. Again, with the Lillihei-Kaster inlet valve in a tangential inflow tract, lift of the ball valve was noted.

**Free and Forced Vortex Flow Fields**

Throughout this section, references have been made to two-dimensional vortical flows within the pump. In actuality, these flows may be more adequately described as helical flows in a three-dimensional sense. However, for mathematical simplicity, only two-dimensional flows will be considered.

Inflow into the pump was speculated to be a poorly developed forced vortex, i.e., rotational flow. By definition of a forced vortex, the velocity of the particles closest to the wall are higher than those at the center of the pump. The velocity distribution of this type of flow may be seen in Figure 3.25. Motion to the inner portion of the pump's fluid was imparted by viscous forces unless previous rotational flow existed.

Upon ejection of the pump's contents, a velocity distribution more like a free vortex was established as shown in Figure 3.26. This type of idealized flow was irrotational except at the center of the vortex where all vorticity was concentrated. In such flows, the vorticity is defined as:

\[
\bar{\omega} = \nabla \times \bar{V} = 2\bar{\omega}_p = 0
\]
FIGURE 3.25

Forced Vortex

Rotational Flow

FIGURE 3.26

Free Vortex

Irrotational Flow
where:

\[ \bar{\omega} = \text{vorticity} \]

\[ v \times \bar{V} = \text{curl of the velocity} \]

\[ \bar{\omega}_p = \text{angular velocity} \]

In rotational flow each infinitesimal particle in the flow field rotates about its own axis, while in irrotational flow an infinitesimal particle does not rotate about its axis. The mean rotation is equal to:

\[ \frac{1}{2} \left( \frac{\partial v}{\partial x} - \frac{\partial u}{\partial y} \right) \]
If these two types of vortices were formed during the pump's cycle, they would disperse any stagnating residual columns of fluid in the pumping chamber. Further experimental evidence of this conclusion is presented in a later section. Summary information and formulation of a new pump design will be considered in the next chapter.
CHAPTER IV

Design Considerations of the New LVBP

In redesigning the LVBP, the following observations were made:

(1) The tangential inflow duct with a flow directing, tilting disc valve provided the best "washing" of the pump's inner surfaces.

(2) The angle of the inlet to the pump must be set in such a way as to direct inflow toward the diaphragm.

(3) The diaphragm should not come in contact with the body surface of the pump or with the protruding supportive structures of the inlet valve.

(4) The distance through which the diaphragm should travel, i.e., its axial displacement, should be kept to a minimum to avoid high stress concentrations developed in its wrinkled surface, and to avoid pump wall-valve contact.

(5) The outlet valve should be a tilting disc valve. This conclusion was based on the process of eliminating the ball and the occluder disc valves because of their undesirable flow characteristics. These included valve noise and susceptibility of supportive structures to flow stagnation and thromboembolic complications.
(6) The diameter of the outflow was made larger to decrease the angular velocity of the outflowing vortex.

(7) The kinetic energy of the pumping system is dependent upon the diameter of the pump, the total volume, and the inlet flow velocity. If one selected a smaller axial diaphragm displacement with the same stroke volume, a larger diameter pump would be required. Assuming the same inlet velocity, the kinetic energy of the pump fluid of the larger diameter pump would be greater. The tendency for pumping chamber flow stagnation during decreased pulse rates, such as when the patient must be weaned from the pump, would also be decreased.

A cutting plane drawing and photographs of the redesigned pump are shown in Figures 4.1 and 4.2(a) and (b). This pump incorporated the above seven modifications in order to improve internal pump flow characteristics. Analysis of the flow visualization films of this pump geometry is contained in the next section. Additional experiments were devised to test the new pump design.

**Redesigned LVBP Flow Studies**

In this new bypass pump arrangement, the inlet valve was a small Lillihei-Kaster, while the outlet valve was a large Lillihei-Kaster. The inlet duct was tangential to the pump body cone and was angled in such a way as to direct flow
toward the diaphragm region of the pump (Figure 4.1). The low profile diaphragm was constructed from silastic and displaced approximately 60 cc of fluid. The total pump volume was 250 cc while the total axial diaphragm displacement was 1.25 cm.

In the film showing the operating bypass pump, no areas of stasis or flow stagnation were apparent. Swirl or vortical flow within the pump body was the prominent flow feature. Figure 4.3 shows flow in a pump cross section at the inlet connector. Note the parallel streaks established by the swirling motion during pump inflow. Figure 4.4 shows the same inflow in a view from the underside of the pump. Figure 4.5 illustrates the backflow associated with a closing outlet Lillihei-Kaster valve. Note also the non-occlusive end of travel position of the diaphragm and the vortical swirl established around the pump's vertical axis.

Fluid entrapment between the diaphragm and the pump body did not occur because of the relatively short axial displacement of the diaphragm. Diaphragm contact with the inlet valve support structure was also eliminated.

This redesigned bypass pump was approximately 3.50 inches in diameter as compared to 2.25 inches in the old bypass pumps. If one assumed the same inlet velocity for both pumps, i.e., the same diastolic duration and filling volume, the kinetic energy associated with the larger pump would be greater. This increase in energy would be desirable if pump
pulse rates and flow rates were reduced during a patient's "weaning" period. Observing pump internal flow under pulse rates of 30 bpm illustrated that the swirl "washing" of the pump body was maintained during diastolic periods approaching 1.5 seconds.

Because of the positioning of the inlet valve in the inflow duct, eddy formation on the distal side of the inlet valve was seen during pump systole (Figure 4.6). This vortex formation was caused by the square corner associated with the placement of the inlet valve with the pump housing.

Figures 4.7 and 4.8 are views of an operating pump looking into the inlet valve. Note the swirling flow both during pump diastole (Figure 4.7) and pump systole (Figure 4.8).

Another film of this new pump design was made using a large Braunwald-Cutter ball valve in the outflow position. The flow directing inlet Lillihei-Kaster valve established swirling flow within the pump. The ball valve again appeared to be unstable in the outflow tract. The ball valve only briefly reached a full open position during systole and oscillated near a seated position during the remainder of the pump's ejection period. Flow stagnation on the distal of the valve and on its cage was again apparent due to the vortical nature of the outflow.
FIGURE 4.6

FIGURE 4.7
Stagnation-Dye Flow Studies in Pulsatile Pumps

A brief study was conducted in order to determine if a stagnating column of fluid was established by the vortical flows within the pumps. Both the old and new bypass pumps were equipped with silastic diaphragms to which flexible silastic capillary tubing was attached at the diaphragm's apex (Figure 4.9). This tubing was directed from the pump through the air driveline into a plastic "Y" connector. At this point, a rubber seal with a hyperdermic needle was attached to the tubing. To this needle, an India ink-filled syringe was connected. To the remaining half of the "Y", the air driveline was continued to the pneumatic pulse unit.

Cine films of this experiment were made at 180 frames per second using a 1000 watt flood light source. Dye was periodically injected into the pump via the capillary tubing. This capillary tubing did not limit the normal movement of the diaphragm.

The results of these films were encouraging. The dye injected slowly into the center of the flow was dispersed throughout the pump's volume during pump diastole. No core stagnation regions were established within the pumps. The ink in the old pump was cleared more quickly than the new LVBP because of its smaller residual volume. Also there appeared to be no radial dispersion of the ink caused by its
initial injection velocity. The only dispersion occurred because of velocity changes and mixing within the pump.
CHAPTER V

Conclusions

In this thesis, a flow visualization method was implemented in the study of the fluid flow characteristics of a swirl type left ventricular bypass pump. A systematic combination of pump spatial inlet orientations, pump valves, and pump body geometry were examined in order to determine good pump flow characteristics. Both still and high speed motion photography were made of operating transparent pumps in a mock circulatory loop. Regions of pump stasis, low velocity flows, and eddy formation were identified.

In redesigning the LVBP, several changes were implemented that stemmed from these flow visualization studies. The final pump design incorporated an angled, tangential inlet duct with a small tilting disc valve and a large tilting disc valve in the outflow tract. A low profile, non-occlusive diaphragm was used with a nozzle shaped pump body geometry to improve internal pump flow. These modifications should minimize areas of possible clot formation and fibrin deposition within the pump due to low velocity flows.

Reviewed below are some of the conclusions reached in this study:
(1) The tangential inflow duct with a flow directing, tilting disc valve provided the best "washing" of the pump's inner surfaces.

(2) The angle of the inlet to the pump must be set in such a way as to direct inflow toward the diaphragm.

(3) The diaphragm should not come in contact with the body surface of the pump or with the supportive structures of the inlet valve.

(4) The distance through which the diaphragm should travel, i.e., its axial displacement, should be kept to a minimum to avoid high stress concentrations developed in its wrinkled surface, and to avoid diaphragm, pump wall-valve contact.

(5) The outlet valve should be a tilting or hinged disc valve.

(6) The diameter of the outflow was made larger to decrease the angular velocity of the outflowing vortex.

(7) Corners associated with the inlet connector or with the outlet connector or its valve should be minimized.

(8) Pump valve stability required a continuously moving diaphragm; avoiding large end travel oscillations.

The ultimate test of such bypass pumps are in vivo experiments. These experiments are fraught with problems including surgical error, pumping control, pump biomaterials and animal pharmalogical management. In evaluating
experimentally used pumps, a series of in vivo tests should be made with identical bypass pumps to determine pump inadequacies.

In examining this particular type of bypass pump further, one should examine the possibility of using inexpensive valves in this device, e.g., a flap valve in the inlet position.

All the pumps used in this study were symmetric with respect to the pump's vertical axis. An asymmetric pump body geometry, perhaps of a ellipsoidal cone should be investigated.
BIBLIOGRAPHY


