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Title of Thesis — Titre de la thèse

~~VAN~~ VALVE ORIENTATION EFFECT ON BLOOD FLOW DISTRIBUTION

University — Université

U. OF ALBERTA

Degree for which thesis was presented — Grade pour lequel cette thèse fut présentée

M. Sc.

Year this degree conferred — Année d'obtention de ce grade	Name of Supervisor — Nom du directeur de thèse
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Valve Orientation Effect on Blood Flow Distribution

by

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A THESIS

SUBMITTED TO THE FACULTY OF GRADUATE STUDIES AND RESEARCH

IN PARTIAL FULFILMENT OF THE REQUIREMENTS FOR THE DEGREE

OF Master of Science

Department of Mechanical Engineering

EDMONTON, ALBERTA

Spring 1984

THE UNIVERSITY OF ALBERTA

RELEASE FORM

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Date 26th Oct 1983

ABSTRACT

The main objective of this study was to investigate the effect of the angular orientation of the prosthetic aortic valve of the tilting disc variety on the arterial flow distribution downstream of the valve. For this purpose, a full-scale, elastic model of the arterial system was manufactured; the system included the arterial tree from the heart to the iliac bifurcation and a pump system simulating the heart's pumping action. The results of these experiments showed no dependence of arterial flow distribution on the aortic valve angle.

As a side study, experiments were also conducted on the effects of arterial wall elasticity on arterial flow distribution. Rigidity of the arterial model was simulated by sealing the model tank to prevent expansion of the water surrounding the model. The results from these experiments showed that the flow rate through the arteries changed from 0.25% to 4.9% due to the increased rigidity.

ACKNOWLEDGEMENTS

The author wishes to thank the following individuals for their assistance in the research leading to the production of this thesis:

* Dr. Cz. M. Rodkiewicz for his supervision of the research and encouragement during the later stages of the project

* the technicians who aided in the minor repairs and modifications of the apparatus

* the machinists who performed the major design and repair work on the apparatus; in particular, Mr. Albert Yuen for his excellent and precise work in the modification of the model left ventricular chamber

* Dr. J. Centkowski for his help in redesigning the apparatus and aid in the experimental procedure

* her brother, Raymond, for his expert advice on the most efficient method of producing this thesis on the computer

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NOMENCLATURE

The following abbreviations were used for the naming of the arteries in the model:

CORO	coronary arteries
ILIA	iliac arteries
LEXC	left external carotid artery
LINC	left internal carotid artery
LSCL	left subclavian artery
LVER	left vertebral artery
RENS	renal and abdominal arteries
REXC	right external carotid artery
RINC	right internal carotid artery
RSCL	right subclavian artery
RVER	right vertebral artery

CHAPTER I

INTRODUCTION

The development of prosthetic heart valves arose out of the attempts by physicians to alleviate the problems of valvular stenosis (narrowing of the aortic valve region) and regurgitation (backflow through the aortic valve). Prior to the 1950's, the surgical procedures included valvular commissurotomy (enlarging the valvular orifice physically by the use of a valvulotome or finger) and valvular annuloplasty (constricting the size of the valve by placing sutures around the outside curcumference and tightening the ring) [1]. However, these procedures achieved limited relief of the original problem and even added further complications due to the surgical procedure.

In 1952, Hufnagal successfully implanted a ball-and-cage prosthetic valve in the descending aorta of a patient suffering from aortic regurgitation. The procedure was a success and led to rapid improvements in the field of prosthetic heart valves. Eight years later, in 1960, Harken performed an open heart operation for implantation of a caged ball prosthesis in the aortic root region.

These and other developments around the world inspired the need for the Conference on Prosthetic Valves for Cardiac Surgery in 1960. The topics at this time concentrated on an understanding of cardiac valve mechanics, surgical techniques and first hand accounts of valve replacements [2]. At the Second Conference on Prosthetic Valves for

Cardiac Surgery in 1968, the improvements in the field were evident. More sophisticated methods for *in vitro* testing of prosthetic valves were detailed; extensive investigations had been done on material bio-compatibility; and valve replacements were being done routinely in both adults and children [3].

Currently, there are three major types of prosthetic aortic valves - ball-and-cage, tilting disc and tissue valves - the first two being mechanical valves and the last a biological valve. Both categories have inherent difficulties. With the mechanical valves, the problem of bio-compatibility necessitates the use of anticoagulation therapy, with its attendant detrimental side effects. On the other hand, tissue valves possess an unproven long-term durability, which may result in re-operation [1].

Most of the research done nowadays concentrates on improving these two aforementioned difficulties. Additionally, work is done to evaluate and improve the *in vitro* (in an artificial environment) and *in vivo* (within the living body) performances of prosthetic heart valves. Tests are done to determine such characteristics as the bio-compatibility of the material, resistance to clotting, tendency to cause blood damage, durability of the prosthesis, pressure drop and energy loss across the valve, and the amount of regurgitation through the valve [4].

Such areas of research focus on the design and performance aspects of the prosthetic heart valve. Another

possibility for study lies in the consequences of positioning the implanted prosthetic valve in different ways. In particular, the effect of the angular orientation of the aortic valve on the arterial flow distribution can be studied.

CHAPTER II

LITERATURE SURVEY

The object of this literature survey is to ascertain the work done by previous researchers on the consequences of varying the angular orientation of the prosthetic aortic valve of the tilting disc category (see Chapter IV, Section iv).

An extensive search of such indices as the Cumulated Index Medicus and the Engineering Index showed the lack of study done on the subject of aortic valve orientation.

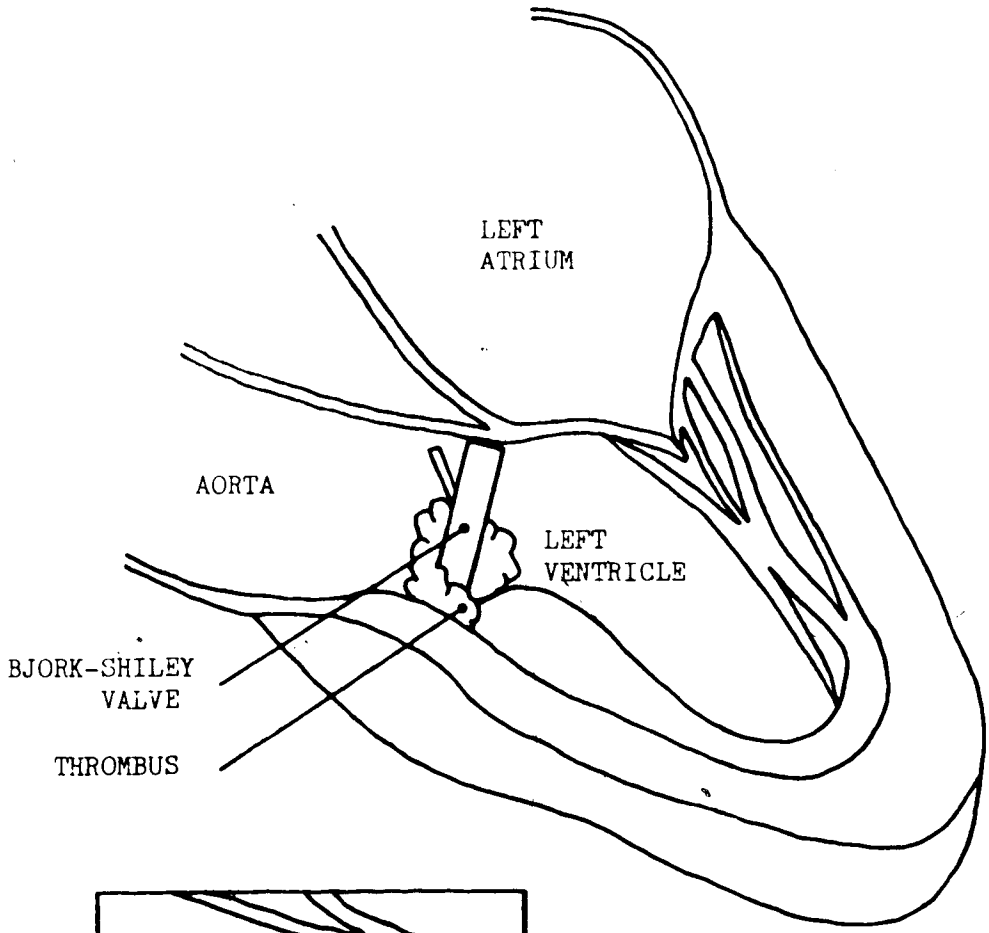
Bozer and Karamehmetoglu [5] reported in 1972 the incidence of thrombosis (clot formation) with the Bjork-Shiley tilting disc artificial heart valve. In particular, thrombi (clot) formation was found on the small aperture of the valve. (Since the disc tilts open to an angle of 60° only, the valve orifice is divided into two apertures, one large and one small. Chapter IV, Section iv gives more details on this subject.) This prompted the authors to speculate that since the volume of blood flow through the small aperture is less than that through the large aperture, there is a region of "relative stasis" around the small aperture which encourages thrombus formation there.

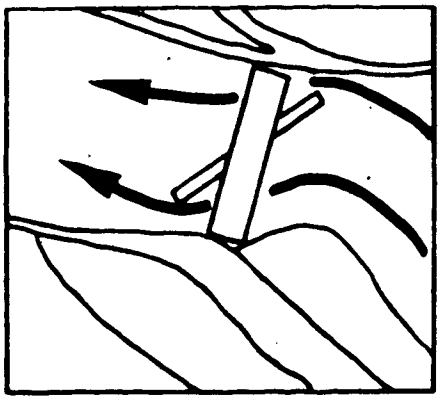
In 1976, Fernandez et al [6] detailed seven case studies of patients who encountered thrombus formation with their implanted Bjork-Shiley aortic valve prosthesis. Case 6 is of particular interest to this study. After

implantation with a Bjork-Shiley prosthesis, this patient encountered problems with the valve. Re-operation indicated a thrombus attached around the smaller aperture of the aortic prosthesis. This was removed and it was seen that the left ventricular wall might have disturbed the blood flow through the smaller opening and contributed to the thrombosis. The valve was rotated 180° such that the smaller aperture faced away from the left ventricular wall (Figure 1). This seemed to solve the problem as the patient is reported to be doing well. This report shows that the orientation of the tilting disc aortic valve does have an effect on the flow characteristics through the valve; in this case, the regions of "disturbance" around the small aperture of the valve were affected.

More recently, in 1980, Sabbah and Stein [7] published a paper on a study showing the effect of aortic valve orientation on the disc opening of two tilting disc valves. The authors showed that the full opening of these valves appears to be affected by the orientation of the valve relative to gravitional forces. This would indicate that the valve orientation may affect the flow characteristics past the valve.

A more specific study on the extent of such an effect was undertaken by Chandran et al [8]. In that study, the effect of aortic valve orientation on the flow development in the aorta was investigated. This was a flow visualization study using neutrally buoyant beads. Two





BJORK-SHILEY VALVE ROTATED
180°, FLOW FLOWS THROUGH
RELATIVELY UNDISTURBED

Figure 1: Interference of left ventricular wall on Bjork-Shiley tilting disc valve action and solution to the problem.

valve angles for the tilting disc valve were tested; with the disc tilting either towards the outer curve of the aorta or towards the inner curve. The results showed the better position to be the former. In this case, the flow was more uniform and there were no stagnation regions present.

Even though this last study [8] approaches closely to the present study, it is not directly applicable, as the flow distribution downstream of the aortic valve is not considered.

It can thus be seen that the subject of the present study has not been undertaken by any researcher yet. However, the aortic valve orientation does have some effect on the flow characteristics around [5 and 6] and past [8] the valve.

CHAPTER III

DETAILS OF PROBLEM

The orientation of a tilting disc valve in the aortic position has been shown to have certain effects on the blood flow patterns downstream of the valve. Among these are the location and form of the stagnation zones immediately downstream of the valve.

The object of the present investigation concentrates on a different consequence of the orientation of the aortic valve. In particular, it is postulated that the blood flow distribution to the arteries leading off the aortic arch may be a function of the aortic valve orientation. To test this theory, one has to resort to experimental methods since construction of a mathematical model of the arterial system, with its variety of branching angles and diameters is an almost impossible task.

However, even an experimental model of the arterial system does not offer the perfect solution. Besides the problems of maintaining geometrical and dynamic similarity, one must also consider the effects of using an artificial heart pump or substituiting a rigid arterial wall instead of one duplicating the natural elasticity of the arterial wall. The latter point has been a controversial issue for some time. Caro et al [9] states that the distensibility of the arterial wall may be neglected when considering the flow rates and velocity profiles in a blood vessel. However, as explained in [10] and detailed later, the elasticity of the

arterial wall does affect the blood flow distribution to the different arteries.

In summary, this study was undertaken to investigate the effect of aortic valve orientation on the blood flow distribution to the different arteries. A supplementary study was also performed to show the importance of including elastic arterial walls in certain experiments on *in vitro* models of the arterial system.

CHAPTER IV

APPARATUS

To carry out the experiments on valve orientation and wall elasticity, an *in vitro* model of the human arterial system was used. This full-scale model duplicates as closely as possible the characteristics and parameters of the corresponding *in vivo* system. For ease of presentation, the model is divided into three subsystems:

i) Heart Pump System

ii) Arterial System

iii) Measuring System

In addition, a section detailing the types of prosthetic aortic valves and the characteristics of each type has been included since the major topic of this study concerns the prosthetic aortic valve.

i) Heart Pump System

The pumping action of the left ventricle of the human heart was duplicated by a cam-piston arrangement, with the pulse transmitted to the working fluid through an impermeable membrane (Figure 2).

The cam was hand-manufactured from a solid Teflon disc, with the groove cut such that the velocity profile generated by the piston stroke approximated that found at the entrance to the aorta *in vivo* (Figure 3). The gear train, to which

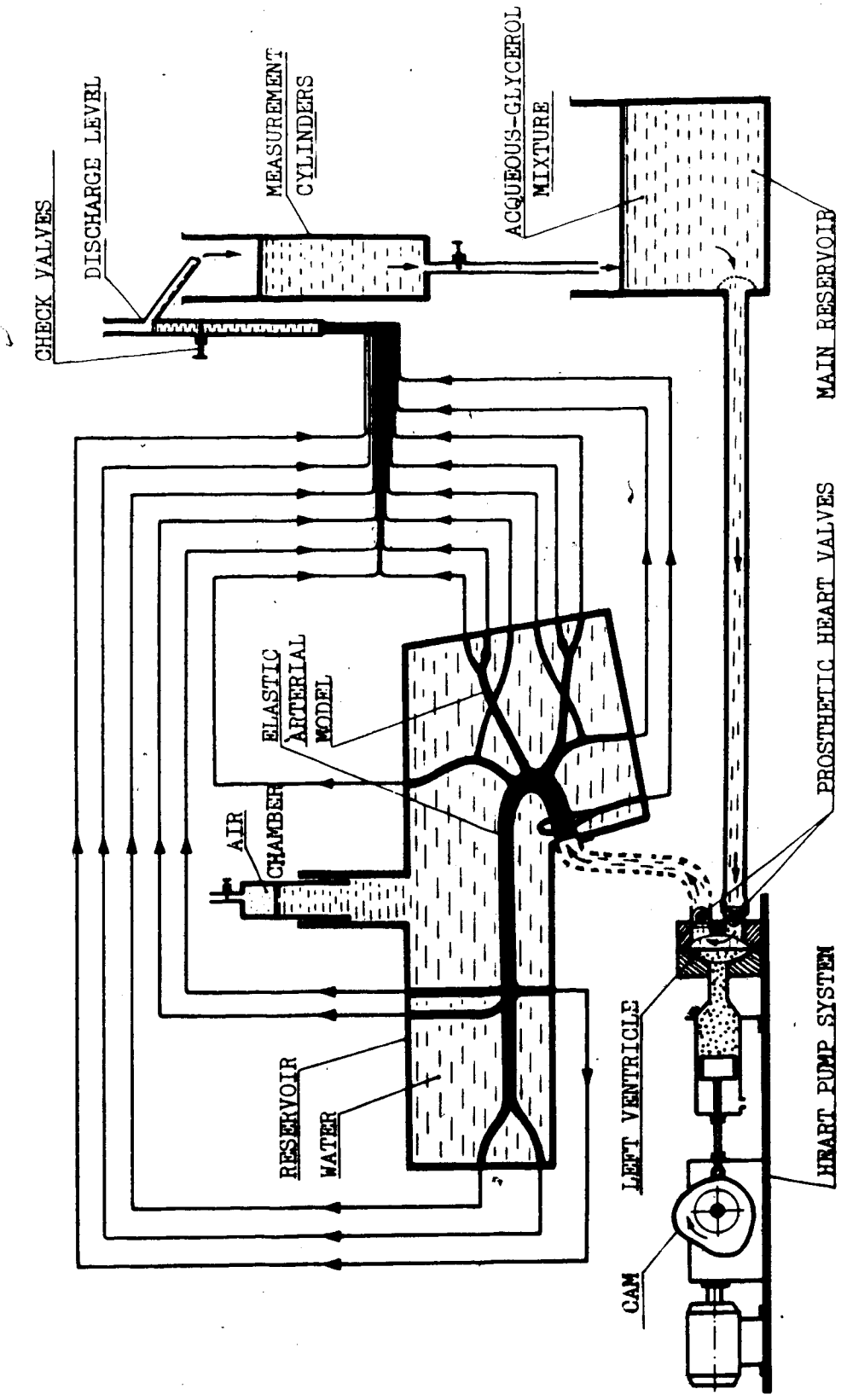


Figure 2: Schematic of apparatus used for the study

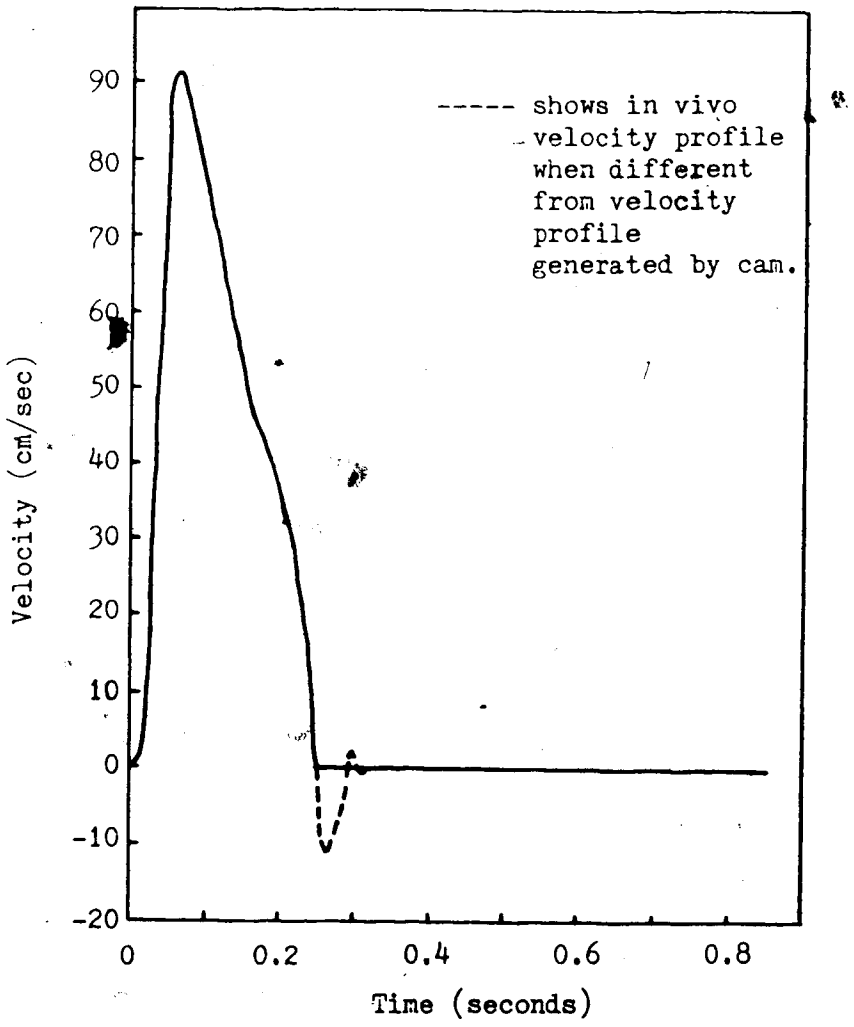


Figure 3: Velocity profile found at the entrance to the aorta in vivo compared to velocity profile generated by cam.

the cam is attached, is interchangeable to allow for three different speeds, resulting in pulse rates of 50, 70 or 90 beats per minute at the piston end. For this study, however, the speed was set for a pulse rate of 70 beats per minute, representing the normal adult human pulse rate.

The piston-cylinder unit was designed for a volume output of 5.679 liters per minute at a pulse rate of 70 beats per minute. This volume output would correspond to the cardiac output *in vivo*. It should be noted, however, that the cardiac output of the model is dependant on a number of factors, namely, the pulse rate, piston liquid volume and type of prosthetic valves used to control the direction of fluid flow in the arterial system. For this study, however, only the last two factors were important, the first one being kept constant throughout the experiments.

It was found that the cardiac output could vary as much as 6% depending on the liquid volume in the piston cylinder. The air chamber on top of the piston cylinder provided the means for varying the piston liquid volume.

The quoted cardiac output of 5.679 liters per minute was one of the original design criteria, and was dependant on the usage of certain prosthetic valves. Specifically, the design called for two Smeloff-Cutter ball valves (see Chapter IV, Section iv), which performed the function of the mitral and aortic valves located in the left half of the heart (Figure 4). However, the main objective of this study

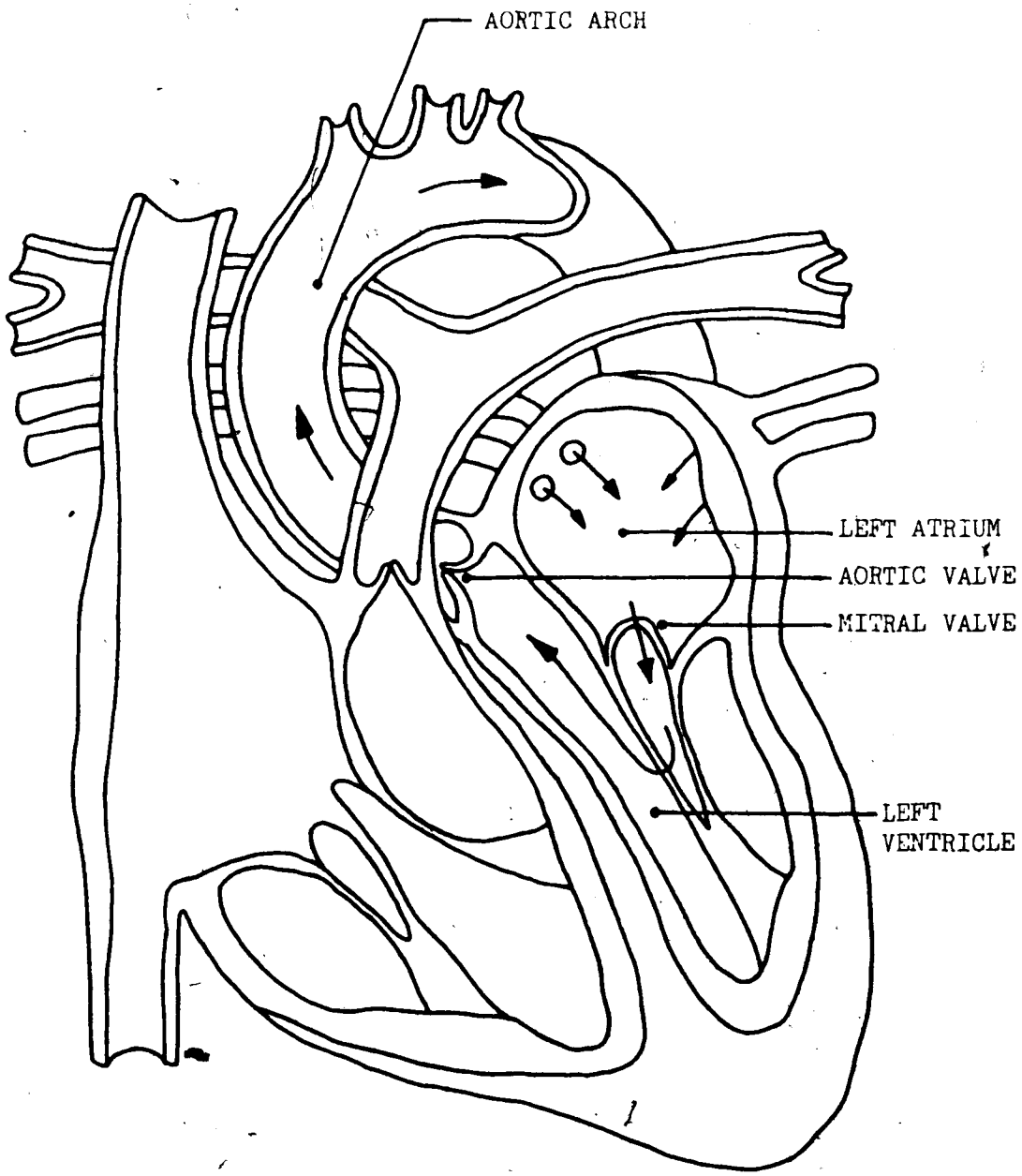


Figure 4: The natural human heart, showing the location of the aortic and mitral valves.

was on valve orientation, and it can be appreciated that ball valves, possessing axial symmetry, are not appropriate for such experiments. Therefore, a Bjork-Shiley tilting disc valve (see Chapter IV, Section iv) was installed in the aortic valve position. This resulted in a cardiac output of 5.934 liters per minute. However, this difference was not deemed important since the flow distribution was the analysis criterion.

The impermeable diaphragm, installed in the egg-shaped chamber representing the left heart, served to separate the piston liquid, distilled water, from the working fluid. The diaphragm was cut from a piece of rubber latex, which was found to be durable but still possessing adequate elasticity to transmit faithfully the impulses generated by the piston.

The working fluid was composed of a mixture of 36.7% glycerol with distilled water. This provided a viscosity of 4.6 centipoise, which represents the viscosity of blood in a human being with normal hematocrit (volume percent of red blood cells in whole blood). In addition, the chemical Zephiran Chloride was added at a concentration of 0.133% (1:750). Zephiran Chloride is normally used as a hospital antiseptic for instruments and rooms but in this application, it serves to eliminate any bacterial growth in the model due to the presence of glycerol.

ii) **Arterial System**

The human arterial system was modelled in some detail to include the aortic arch and its branches, the abdominal aorta and the lower abdominal aorta with the iliac bifurcation (Figure 5).

The arteries leading off the aortic arch were duplicated in some detail as the original design called for a thorough study of the cerebral (brain) blood supply [11]. Immediately past the aortic valve, the right and left coronary arteries branch off. These arteries, *in vivo*, supply blood to the muscle tissues of the heart. Just before the maximum curvature of the aortic arch, the brachiocephalic artery branches off. This artery, also known as the innominate artery, in turn gives rise to the right subclavian artery, right vertebral artery and right common carotid artery. The last artery branches once more into the right internal and external carotid arteries.

Approximately at the apex of the aortic arch, the left common carotid artery branches off, dividing later into the left internal and external carotid arteries. The last major branch off the aortic arch is the left subclavian artery, with the left vertebral artery leading off it.

In the human body, the subclavian arteries supply blood to the arms. The external carotid arteries supply blood to all parts of the head except the brain, leaving the internal carotid arteries and vertebral arteries to furnish the cerebral blood supply.

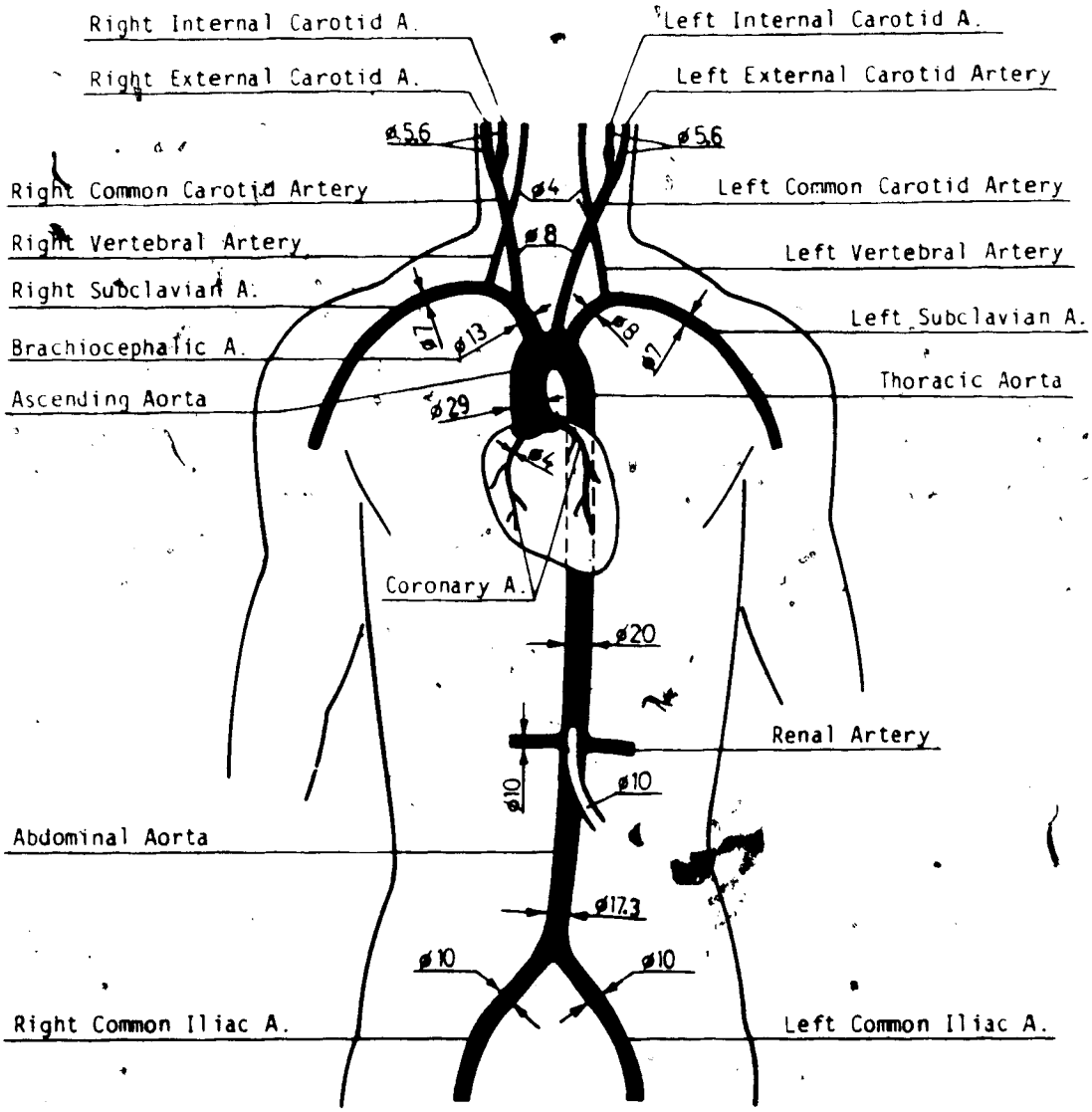


Figure 5: Schematic of the arterial system modelled for the study, from [11].

Once past the last branch, the aorta is generally known as the thoracic aorta, possessing an upper and lower portion. From the thoracic aorta branch the intercostal arteries, which supply blood to the chest region. These arteries were not modelled as the amount of blood flowing through them was considered too little to affect the overall flow through the model arterial system. Once the aorta passes through the diaphragm separating the chest and abdominal cavities, it becomes known as the abdominal aorta. The major branches in this region are the renal arteries and mesenteric arteries. The former transport blood through the kidneys, which perform the vital task of removing impurities from the blood while the latter supply blood to the digestive organs of the abdomen, for the all-important task of digestion. The mesenteric arteries and other minor abdominal arteries were combined into one artery in the model.

Past the abdomen, the aorta terminates at the iliac bifurcation, where the left and right iliac arteries branch off. The iliac arteries in turn branch repeatedly, supplying blood to the legs and lower abdominal region.

The abovementioned arteries constitute the arteries modelled for this study. A more complete picture of the model, including the dimensions can be gained from Figure 5. It should be noted here that the given description of the arterial system is an over-simplified one and while it gives one an understanding of the functions of the various

arteries, it is by no means an exhaustive or detailed account. For such, one must consult a medical anatomy text book such as Gray's Anatomy listed in the Bibliography.

The diameters (Figure 5), branching angles and normal flow distribution (given in Figure 6) were all obtained from a thorough search of the literature. As all human beings are not identical, the values obtained and incorporated in the design are not universal but were chosen to represent the majority of human beings.

The fabrication of the arterial model consisted of two parts. First, solid molds representing the inner dimensions of the arteries were manufactured. The model was broken down into smaller sections for ease of fabrication. Each arterial section was machined to a smooth finish and then polished as finely as possible. This ensured the smoothness of the inner wall of the model arteries. The molds were made either of plexiglass or stainless steel.

The next step of the fabrication process consisted of applying successive coats of a liquid mixture of silicone and toluene. Each layer was applied by immersing the mold in the mixture, after which the coated mold was clamped to a drill and rotated slowly. The rotation aids in the drying process and also spreads the mixture evenly around the mold. In this way, as many as 20 coats of silicone and toluene are applied, the exact number depending on the shape of the mold and the artery thickness desired. Complete details may be obtained from [11]. After a curing period of one day, the

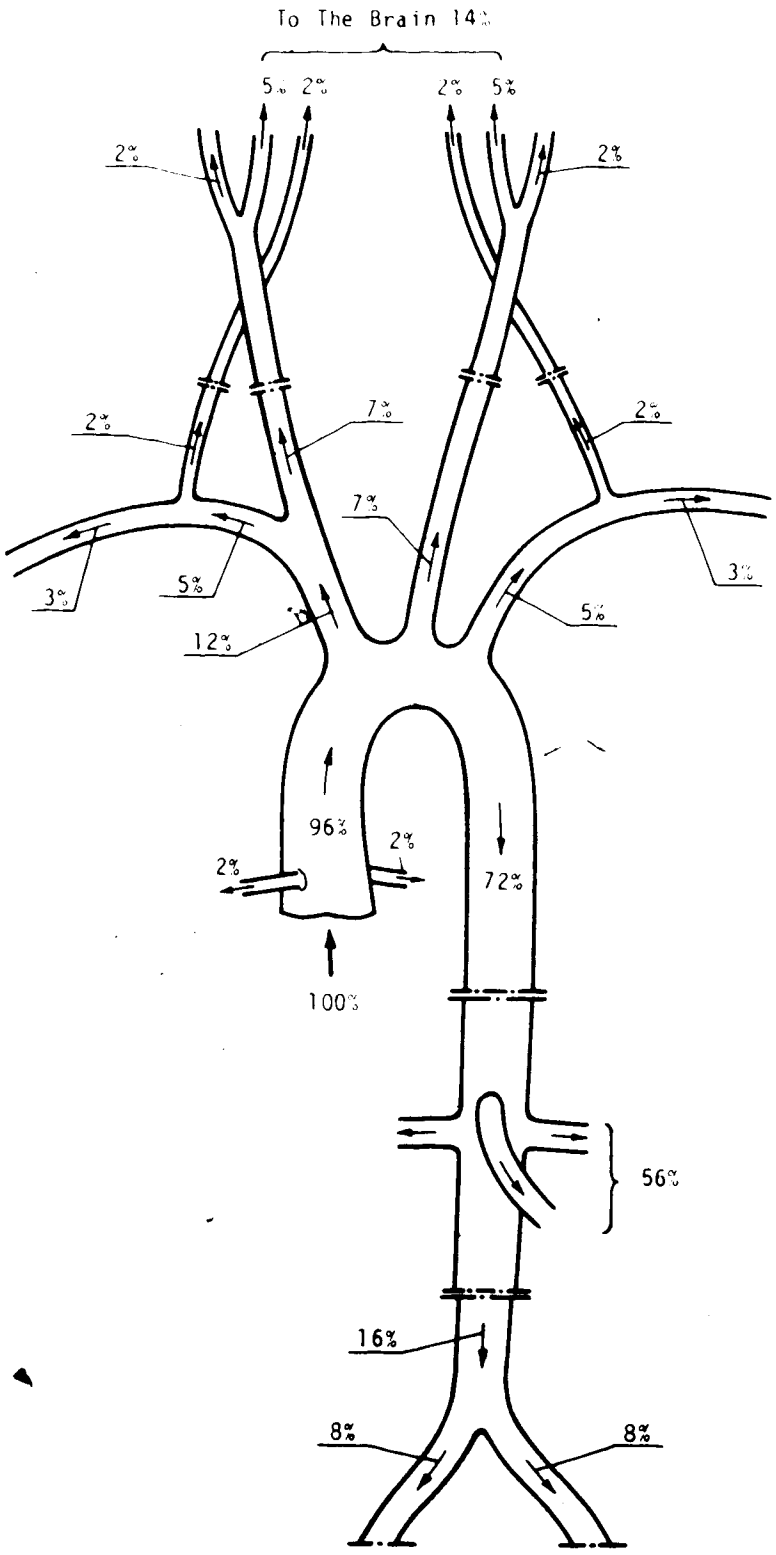


Figure 6: Normal flow distribution for arterial system, from [11].

silicone models were removed from the molds and cured for a further period of about one week.

Considering the manufacturing process, the model arteries are composed almost entirely of silicone since most of the toluene evaporates during the drying process. Thus, the walls of the arteries possess a certain degree of elasticity.

After all the arterial parts were manufactured, the whole model was assembled using brass connectors having identical internal diameters as the respective arteries attached. The model was then installed in a horizontal position in a large plexiglass tank. This corresponds to the resting position of the human body. The model was held in position by wires suspended from rods placed across the top of the tank. The discharge end of each artery was attached to connectors on the walls of the tank. From the outer end of these connectors, Tygon tubes was attached, leading to the measurement cylinders.

To simulate the properties of the surrounding tissue *in vivo*, the plexiglass tank was filled completely with water. Furthermore, the tank was sealed and a slight pressure applied by means of an air chamber attached above the tank (Figure 2).

This air chamber was an important component of the experiments on the elasticity of the arterial walls. When left open to the atmosphere, the arterial model was permitted to expand and contract with each pulse. However,

when this chamber was filled with water and sealed, the expansion and contraction of the arterial model was drastically reduced due to the virtual incompressibility of water at ambient temperatures and pressures. The model was not rendered completely rigid, though, since the walls and lid of the model tank did move with each pulse. This was somewhat minimized by the placement of weights on top of the model tank lid. Furthermore, some seepage of the tank water occurred through the joints of the model tank. This effect was quite small and no attempt was made to correct it.

iii) Measuring System

Once the working fluid leaves the arterial model, it travels via a set of Tygon tubes to the measurement cylinders. These were placed in one row for ease of data recording (Figure 2). There were 11 measurement cylinders in all, which measured the volume of flow from 11 arteries or sets of arteries, namely, the coronary arteries, right subclavian artery, right vertebral artery, right external carotid artery, right internal carotid artery, left internal carotid artery, left external carotid artery, left vertebral artery, left subclavian artery, renal and abdominal arteries, and iliac arteries. The size and scale for each of these measurement cylinders was such that the maximum resolution was obtained for each cylinder.

The discharge into the measurement cylinders was located about 100 cm above the level of the model. This provided part of the systemic resistance for the model

arterial system, representing the diastolic pressure in the body of 80 mm of mercury. The rest of the systemic resistance was supplied by check valves located in the discharge tube to the measurement cylinders. These check valves could be adjusted individually, to yield the normal volume flow rate through each artery or set of arteries.

Each measurement cylinder could be emptied, by means of toggle valves located at the bottom of each cylinder, into a common reservoir located below the bank of measurement cylinders. This reservoir, in turn, discharged into the inlet of the arterial model. From here, the working fluid passes through the first prosthetic valve, representing the mitral valve, into the egg-shaped chamber and through the second prosthetic valve, representing the aortic valve. After the aortic valve, the working fluid passes through the arterial model, eventually ending up in the measurement cylinders and reservoir, completing the closed loop for the working fluid.

iv) Prosthetic Heart Valves

The natural human heart with the positions of the valves of the left heart is shown in Figure 4. As mentioned in Chapter IV, Section i, these are replaced in the experimental model by the Smeloff-Cutter ball valve and the Bjork-Shiley tilting disc valve.

The natural aortic valve can best be described as a passive three-leaflet valve, the leaflets opening and closing according to the transvalvular pressure difference.

As the left ventricle contracts at the beginning of the systolic (or ejection) phase, the left ventricular blood pressure increases until it reaches the blood pressure in the aorta of 80 mm of mercury. This corresponds to the diastolic (or resting) blood pressure. The ventricular blood pressure then increases beyond the diastolic pressure, forcing the aortic leaflets open. Blood from the left ventricle is ejected rapidly into the aorta, causing a decrease in left ventricular pressure, while the blood pressure in the aorta increases. The reversal of pressure difference causes the aortic valve leaflets to snap shut. Almost simultaneously, the mitral valve opens, filling the left ventricle with blood. In the aorta, the blood moves through the branch arteries, causing a decrease in blood pressure to the resting pressure of 80 mm of mercury. With the increase in left ventricular pressure, the aortic valve opens, repeating the cycle once more.

Failure of the natural aortic valve occurs when the leaflets adhere to each other and do not open sufficiently, or when the leaflets do not close tightly, allowing blood to flow back into the left ventricle. The former case is commonly called aortic valvular stenosis while the latter is termed aortic regurgitation. Both these problems can be caused by infection. The solution, in most cases, is to replace the defective aortic valve with a prosthetic aortic valve. This surgical procedure involves removing the diseased aortic valve and suturing (sewing) the prosthetic

aortic valve in place.

The two most common types of prosthetic valves are the ball-in-cage and tilting disc valves. Other types are the caged disc and tissue valves. Since the experiments of this study involve use of a Smeloff-Cutter ball-in-cage valve and a Bjork-Shiley tilting disc valve (Figures 7 and 8), this discussion will concentrate mainly on them.

The Smeloff-Cutter valve consists of a solid silicone rubber ball enclosed in a double cage made from one piece of commercially pure titanium [12]. The suture ring is made of Teflon cloth. The double cage design allows a larger orifice diameter, equal to the diameter of the ball. According to the manufacturer, this decreases the pressure gradient across the valve and also eliminates regurgitation, because the ball seats precisely in the orifice upon closing. The open-ended cage design is reported to reduce thrombus formation in the wake of the valve.

The Smeloff-Cutter ball valve design is not unique. Another common variation is the single cage design, where the orifice diameter is smaller than the ball diameter. The Starr-Edwards prosthetic valve is a good representive of this design [1].

The Bjork-Shiley tilting disc valve represents the relatively new type of prosthetic valve, having been first designed and manufactured in the late 1960's. (The ball valve design was first conceived in the late 1950's.) With the tilting disc valve, a freely-floating disc is contained

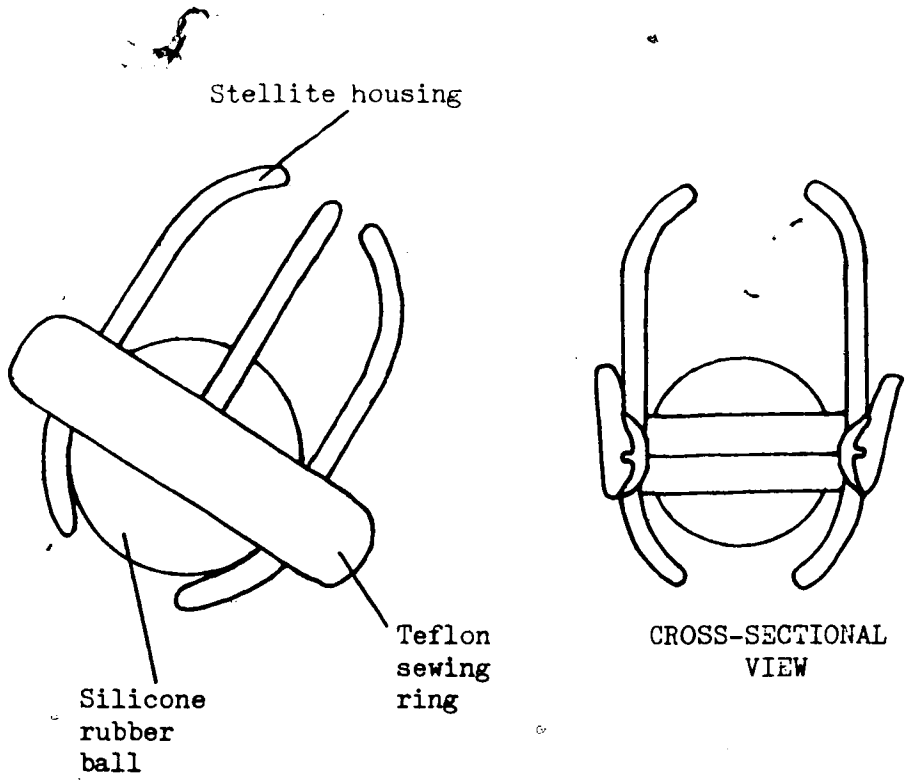


Figure 7: The Smeloff-Cutter ball-in-cage prosthetic heart valve.

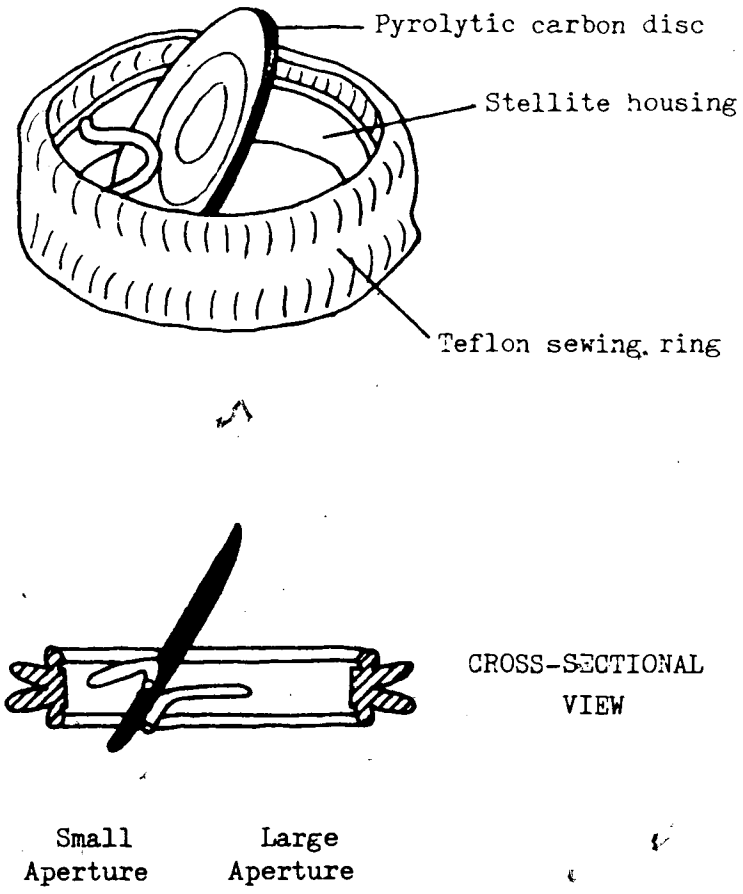


Figure 8: The Bjork-Shiley tilting disc prosthetic heart valve.

within an outer housing which also restricts the motion of the disc (Figure 8).

The Bjork-Shiley tilting disc valve was chosen for these studies on valve orientation mainly because it is one of the original tilting disc valve designs and has not undergone any major changes in design since then. In particular, the model ABP valve was tested. This valve is composed of a pyrolytic carbon disc enclosed within a Stellite housing. The suture ring is made of Teflon and is thin in comparison to that for a ball-in-cage valve. This is one of the advantages of the tilting disc valve compared with other types of valves: the high orifice to suture ring diameter ratio. Other reported advantages include the central flow design, faster response time due to the light weight of the disc and reduced wear on the disc due to its freely-floating nature [13].

Other prosthetic valves under the tilting disc valve category include the Lillehei-Kaster valve, Omniscience valve and Medtronic-Hall valve.

As mentioned in Chapter IV, Section i, the Smeloff-Cutter prosthetic valve was placed in the inlet of the egg-shaped chamber, corresponding to the mitral valve leading into the left ventricular chamber. In the aortic valve position, the Bjork-Shiley ABP prosthetic valve was placed. This section was specially designed to enable the valve to be rotated externally.

The suture ring of the prosthetic valve was removed and the rest of the valve was glued into a specially designed brass ring. Epoxy was used for permanency and great care was taken to avoid any contact between the epoxy and disc of the valve. Next, the valve was attached to the outlet tube of the egg-shaped chamber by means of two screws. This outlet tube was designed to rotate within a connector attached to the tank holding the arterial model. The interior end of this connector was attached to the arterial model. The outlet tube was inscribed with degree markings to indicate the orientation of the prosthetic valve. It should be noted that the rotation mentioned is about an axis perpendicular to the plane of the valve housing. The zero reference position was defined such that the tilting axis of the disc was horizontal and the disc remained open when there was no flow through the valve. The choice of such a reference is, of course, arbitrary, but it can be seen that it is unique.

The attachment of the Smeloff-Cutter prosthetic valve was much simpler since it was only necessary to place it securely in the inlet tube to the egg-shaped chamber. This was done by removing the suture ring of the valve and machining threads into the metal ring. The valve could then be screwed into matching threads on the inside surface of the inlet tube.

CHAPTER V

PROCEDURE

Although the main study was on the effect of prosthetic valve orientation on the blood flow distribution in the arterial system, a peripheral study into the effects of arterial wall elasticity was also undertaken. It would be most logical, therefore, to divide this chapter into two sections, namely:

i) Valve Orientation Experiments

ii) Wall Elasticity Experiments

The latter section includes the procedure for experiments performed on the apparatus described in Chapter IV as well as a brief account of a numerical study done on previously obtained data.

i) Valve Orientation Experiments

As mentioned in Chapter IV, Section i, the heart rate and cardiac output for the model arterial system were set at 70 beats per minute and 5.934 liters per minute respectively. In fact, the only variable in these experiments was the angular orientation of the Bjork-Shiley tilting disc valve.

Before any data were recorded on the effect of the aortic valve orientation, the arterial model was calibrated to yield the normal flow distribution (Figure 6) for a valve angle of zero degrees. This was accomplished by adjusting

the check valves located in the tubes leading to the measurement cylinders. It was found that the volume flow rates in the output arteries were interdependant. This necessitated the use of an iterative process for calibrating the flow rates in the arteries. Furthermore, the flow rates in the arteries leading off the aortic arch were highly dependant on the position of the respective check valves. Thus, any adjustment of the check valves had to be done in small increments. This resulted in a large number of iterations and in fact, the normal flow distribution was obtained only after 60 iterations.

After the arterial model was calibrated, the experiments on the aortic valve orientation were performed. In order to ensure uniformity of ambient conditions, all the experiments were done in one day.

Starting at a valve angle of 0°, the aortic valve was rotated in increments of 15° until a complete revolution of 360° was obtained. A reading of flow distribution was also recorded at a valve angle of 360° to provide a check on any variation in the normal flow distribution at 0°.

The mechanics of each experimental or calibration run were straightforward. For an experimental run, the aortic valve was first rotated to the angle desired. The pump was then switched on, actuating the cam-piston system. This pumping action forced the working fluid through the arterial model and into the tubes leading to the measurement cylinders. It should be noted here that the fluid level in

the entire model at rest was at the level of the fluid in the reservoir. Since the reservoir was located above the arterial model, the arterial model was always filled with the working fluid, with the resting level located in the Tygon tubing leading from the arteries. Thus, when the pump is switched on, the working fluid must first be pumped up to the discharge into the measurement cylinders. As stated in Chapter IV, Section iii, this discharge level was approximately 100 cm above the level of the arterial model. After the working fluid was pumped to the discharge level of all the measurement cylinders, the pump was switched off and the outlet for each cylinder was sealed using the toggle valves. The pump was then switched on again and allowed to run for two minutes. The fluid volume in each of the measurement cylinders was recorded and the flow distribution obtained using simple arithmetic. The measurement cylinders were then emptied and the whole process repeated for the next valve angle.

For the calibration runs, the process is identical except at the end, when instead of repositioning the aortic valve, the check valves located in the discharge tube are adjusted according to the results of the previous run.

After all the flow distribution data had been recorded for all the aortic valve angles, the results were plotted to show the dependence of the flow rate (or volume percent flow) of each branch on the angle of the aortic valve. Only the results for the arteries leading off the aortic arch

were graphed. For the remaining two sets of arteries, namely the renal arteries and iliac arteries, the resolution on their respective measurement cylinders was too low to yield accurate results, and the results were disregarded.

ii) **Wall Elasticity Experiments**

For this set of experiments, the tilting disc valve angle was maintained at 0°. The variable was the elasticity of the arterial walls.

In the first set of experiments, the arterial walls were "left elastic", that is, they were allowed to expand and contract with each pulsation. Following the procedure given in Chapter V, Section i, three sets of flow distribution data were obtained for this condition.

Next, the arterial model was rendered "rigid", as explained in Chapter IV, Section ii. Again, three data sets were obtained.

The results were tabulated for each artery, with the percent difference between the elastic and rigid cases calculated as:

$$\% \text{ difference} = \frac{\% \text{ flow, rigid} - \% \text{ flow, elastic}}{\% \text{ flow, elastic}}$$

As was mentioned in the introduction to this chapter, a brief account will be given on the numerical work done on the effect of arterial wall elasticity on blood flow distribution. A comprehensive report can be found in [14].

For this portion of the study on wall elasticity, data from two previous studies were utilized [11 and 15].

The first study [15] involved work done on a rigid, full-scale model of the aortic arch and its branches. Various combinations of blockages were placed in the arteries and the resulting flow distribution measured.

With the second study [11], the same arterial model as that used in this study was used. However, three different pulse rates and three different cardiac outputs were used, yielding nine combinations in all. For each combination of pulse rate and cardiac output, various blockages were placed in the arteries leading off the aortic arch. The flow distribution was recorded for each case.

In comparing the data from both these studies, some normalization was performed on the results. This was done to remove the influence of the different cardiac outputs between the two studies. The blockage combinations were then examined to yield those identical combinations suitable for comparison. In all, 23 different blockage combinations were matched between the two studies.

The method of comparison was to calculate the square root of the square of the difference in normalized flow rate between similar arteries for each of the 23 blockage combinations. This can be expressed more precisely by d_{mn}:

$$d_{mn} = \sqrt{\sum_{i=1}^{8} \sum_{j=1}^{23} (Q_{ijm} - Q_{ijn})^2} \qquad \begin{matrix} m = 1, \ldots, 10 \\ n = 1, \ldots, 10 \end{matrix}$$

where Q_{ijm} represents the normalized flow rate through the ith artery for the jth blockage combination and for the mth unit.

This is somewhat similar to the procedure for calculating the distance between two points in a space with 115 dimensions. (This number is arrived at because there are 23 blockage combinations and 5 output arteries, the rigid model possessing only 5 output arteries.) Each of the dimensions corresponds to the normalized flow rate through one artery of one blockage combination. The points in this multi-dimensioned space correspond to the location of each unit in the study. For the elastic model, there are 9 units for the 9 combinations of heart rate and cardiac output. The rigid model constitutes the tenth unit. Thus for each unit, the "distance" to each of the other units was calculated. The sum of these "distances", d_m, give a good indication of the proximity of each unit to all the other units:

$$d_m = \sum_{\substack{n=1 \\ n \neq m}}^{10} d_{mn} \qquad m = 1, \ldots, 10$$

Using these distances, a comparison could be made between the elastic model and the rigid model since it can be assumed that if there was no difference in flow distribution between the two models, the distances calculated should be similar for each of the ten units.

CHAPTER VI

EXPERIMENTAL RESULTS

i) Valve Orientation Experiments

The dependence of arterial blood flow on the orientation of the Bjork-Shiley tilting disc valve is shown in Figure 9. The blood flow through the arteries has been depicted as the percent of total cardiac output.

From repeatability studies done on the model, the experimental error was found to be ± 1.5 %. Thus, for all the arteries depicted, the instantaneous variations in flow rate (or percent flow) are all within the experimental error.

However, there is a slight, steady increase in flow rate as the valve angle is varied from 0 to 360 degrees. Furthermore, the flow rate at 360° does not coincide with that for 0°. This would indicate a unidirectional drift in flow rate caused by the equipment. It would also suggest that the deviation is not due to the variation of the aortic valve angle.

ii) Wall Elasticity Experiments

The results of the experiments on wall elasticity, using the arterial model are given in Table 1.

The "error" shown is the absolute error in percent flow. For example, for the coronary arteries, the reading is 3.98 ± 0.01% of the total arterial flow.

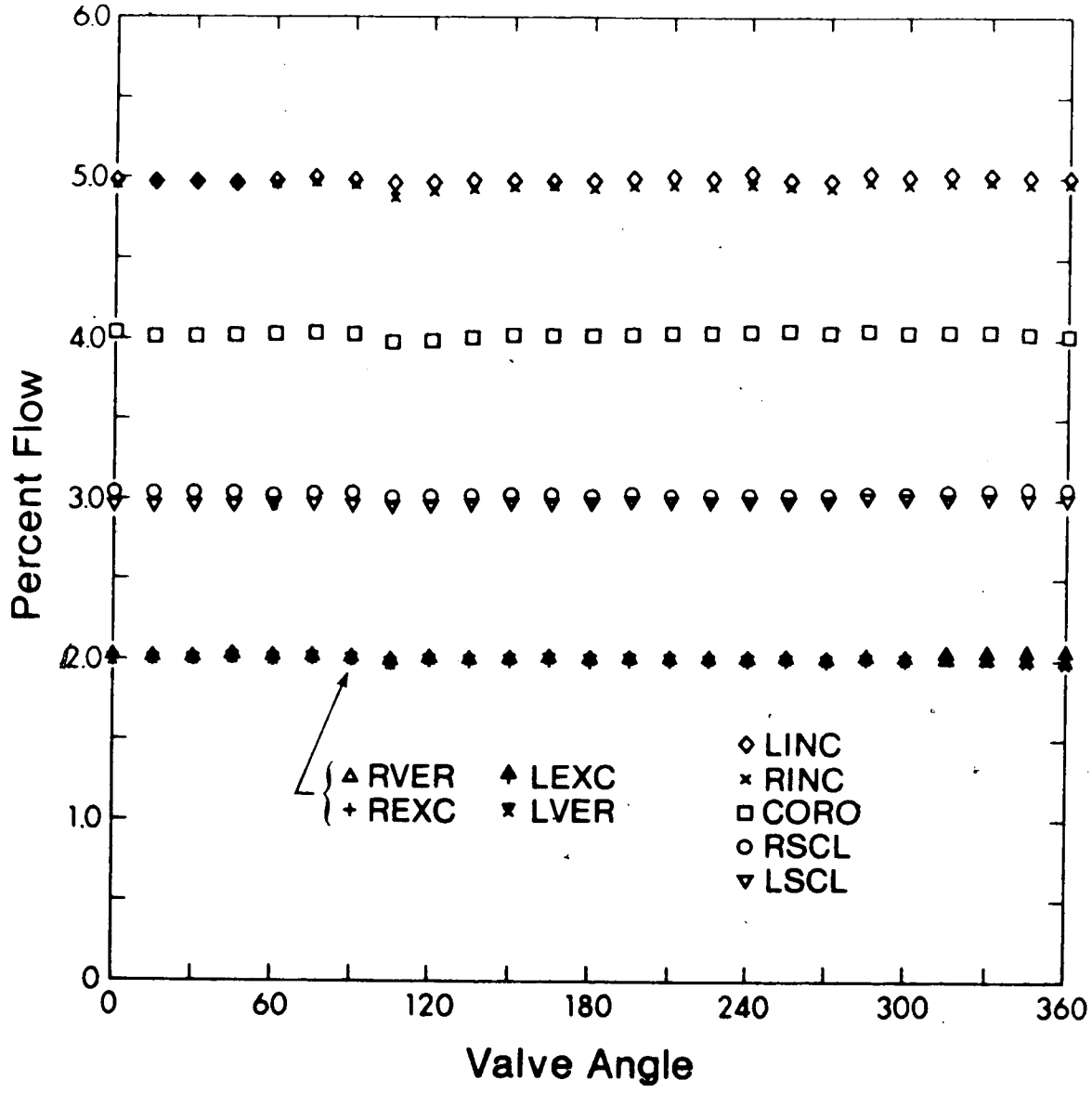


Figure 9: Effect of the orientation of a Bjork-Shiley tilting disc prosthetic heart valve on the flow distribution to the branches of the aortic arch in vitro.

ARTERY	ELASTIC		RIGID		
	%FLOW	ERROR	%FLOW	ERROR	%DIFF
CORO	3.98	0.01	3.97	0.01	-0.25
RSCL	3.01	0.01	3.10	0.01	+3.00
RVER	2.01	0.01	2.09	0.01	+2.70
REXC	2.02	0.02	2.07	0.03	+2.50
RINC	5.03	0.02	5.06	0.01	+0.60
LINC	5.03	0.02	5.06	0.01	+0.60
LEXC	2.00	0.00	2.01	0.00	+0.50
LVER	2.00	0.00	2.03	0.02	+1.00
LSCL	2.97	0.01	3.05	0.01	+2.40
RENS	55.8	0.1	56.1	0.1	+0.54
ILIA	16.2	0.00	15.4	0.1	-4.9

Table 1: Effect of arterial wall elasticity on the flow distribution to the arterial system.

The flow in the two coronary arteries was hardly affected by the increased rigidity of the arterial model. The difference in the flow rate was -0.25%, which is well within the experimental error of 1.5%. With all the other arteries, except the iliac arteries located at the end of the model, the flow rate always increased or was not affected by the increased arterial wall rigidity. The degree of additional flow rate varied from +0.50% to +3.0%, with all values below the experimental error of 1.5% considered insignificant. The largest variation in flow rate was found to occur in the two iliac arteries and amounted to a -4.9% difference, which represents a decrease in flow rate.

The results obtained from the numerical portion of the study on wall elasticity are given in Table 2.

The "distances" between each unit clearly indicate the data from the rigid model to be removed from the data for all units of the elastic model. In fact, from the sum of "distances" for one unit (Table 2b), it is seen that the data for the elastic model form a cluster with the point representing the rigid model outside of this cluster. This shows that there is a difference in the arterial flow distribution between an elastic and rigid model. Unfortunately, the exact magnitude of this difference cannot be ascertained from this kind of numerical study as there are other factors involved besides the arterial wall elasticity. Such factors would include the differences in

(a) Distance between each unit.

	EM U1	EM U2	EM U3	EM U4	EM U5	EM U6	EM U7	EM U8	EM U9	RIG MOD
EM UNIT 1	0									
EM UNIT 2	104	0								
EM UNIT 3	87	44	0							
EM UNIT 4	33	91	75	0						
EM UNIT 5	119	33	65	103	0					
EM UNIT 6	97	41	58	82	32	0				
EM UNIT 7	37	115	93	31	129	107	0			
EM UNIT 8	93	24	43	76	35	31	100	0		
EM UNIT 9	68	57	46	49	67	49	70	40	0	
RIG MODEL	216	233	222	201	229	223	199	218	202	0

(b) Distance from one unit to all remaining units.

EM UNIT 1	855
EM UNIT 2	743
EM UNIT 3	734
EM UNIT 4	740
EM UNIT 5	811
EM UNIT 6	719
EM UNIT 7	880
EM UNIT 8	660
EM UNIT 9	647
RIG MODEL	1942

Table 2: "Distances" between (a) each unit and (b) from one unit to all remaining units obtained from a numerical comparison of the elastic (9 units) and rigid models of the aortic arch.

arterial geometry, aortic arch inlet velocity profiles, working fluid composition, systemic resistance and normal flow distribution between the elastic and rigid models. Of course, the arterial wall elasticity could be the cause of the difference in the data but the proof is beyond the scope of the comparison procedure used here. All the procedure shows is that there is a difference in arterial flow distribution for the elastic and rigid models and this difference could be caused by the change in arterial wall elasticity.

CHAPTER VII

Discussion of Results

i) Valve Orientation Experiments

From Figure 9, and as discussed in Chapter VI, Section i, there is no dependence of arterial flow distribution on the aortic valve angular orientation. This may initially seem unrealistic as it would appear to be intuitively obvious that there should be some effect of a tilting disc valve orientation on the blood flow downstream of the valve.

An important consequence of this finding is that it allows cardiac surgeons to implant a prosthetic aortic valve of the tilting disc variety with any orientation of the disc desired, without fear of affecting the arterial flow distribution downstream of the valve. In fact, without any experimental confirmation, to the author's knowledge, this is the currently accepted procedure. The manufacturer's literature on the Medtronic Hall Prosthetic Heart Valve [16] clearly states that the "aortic valve may be oriented in any position that provides free disc motion." In particular, the surgeon is cautioned to check for disc motion interference due to the aortic wall and suture material, and is permitted to re-orient the valve in any direction if necessary.

This study provides conclusive proof that such a procedure is indeed permissible insofar as the arterial flow distribution is concerned. As mentioned in Chapter II,

there may be other consequences of re-orientation of the aortic valve.

ii) Wall Elasticity Experiments

With reference to the results given in Chapter VI, Section ii), the following possible reasons are given.

The flow rate in the coronary arteries was hardly affected by the arterial wall elasticity. It is possible that since the coronary artery outlets are located right at the entrance to the aortic arch, downstream rigidity characteristics have a minimal effect on the flow rate through these arteries. For the arteries of the aortic arch, no discernible pattern in increased flow rate could be detected. One possible cause of the random increases could be in the elasticity of the model arterial walls themselves, since not all the arteries had uniform thickness (number of coats of silicone-toluene mixture) or identical composition (different batches of silicone-toluene mixture).

The flow rate through the iliac arteries always decreased. This finding is quite logical since in all the other arteries, the flow rate either increases or remains constant. As the iliac arteries are located at the end of the arterial model, the flow rate has to be decreased to compensate for all the upstream increases in flow rate.

Thus, one may conclude that the rigidity of the arterial walls in an *in vitro* model of the arterial system does have an effect on the blood flow distribution. This effect is small, having a maximum value of 4.9% but is,

nevertheless, present. It is also not a consequence of experimental error since there was a definite and consistent difference in the flow rates through most of the arteries.

One result of this finding is that the researcher conducting experiments on the blood flow distribution to the body is obliged to construct an elastic model of the arterial system studied or at least to account for it when presenting results gained from experiments on a rigid arterial model. Thus, depending on the experimental error acceptable, the elasticity of the arterial wall cannot be disregarded when conducting experiments on the arterial flow distribution.

CHAPTER VIII

CONCLUSIONS

It was postulated that the angular orientation of a tilting disc prosthetic heart valve implanted in the aortic valve section of the heart would create downstream effects, in particular, on the arterial flow distribution. To test this postulate, a model of the human arterial system was designed and manufactured (Figure 2). It consisted of a full-scale, elastic model of the arterial tree from the exit from the left ventricle down to the iliac bifurcation and the iliacs themselves. The arteries branching off the aortic arch were duplicated in some detail. The heart's pumping action was simulated by a cam-piston arrangement driven by a motor. The pulse was transmitted to the arterial model by a flexible diaphragm. The working fluid was a mixture of glycerol and water, approximating the viscosity of normal human blood. The flow rate through each of the arteries was measured by volume collection over a fixed time.

Experiments performed on a Bjork-Shiley tilting disc valve have shown the postulate to be untrue. The angular orientation of a tilting disc valve has no effect on the arterial flow distribution downstream of the valve. This result implies that the angular orientation of the prosthetic aortic valve of the tilting disc variety can be disregarded, in relation to arterial flow distribution, when implanting in the heart.

As a supplementary study, an investigation was undertaken to study the effects, if any, of the arterial wall elasticity on the arterial flow distribution. The same apparatus used for the valve orientation experiments was utilized for these experiments on arterial wall elasticity. To simulate rigidity, the air vent (Figure 2) located atop the arterial model tank was filled with water and sealed.

The results obtained showed the arterial flow distribution to be dependant on the arterial wall elasticity. The largest difference in arterial flow rate was found to be 4.9%. This finding implies that, depending on the tolerable experimental error, one cannot ignore the characteristic of arterial wall elasticity when conducting experiments on an *in vitro* rigid model of the arterial system.

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