DYNAMIC BEHAVIOR OF PROSTHETIC HEART VALVES: EVALUATION IN RIGID VS. ELASTIC CONDUITS

Sanaa A. Abdallah* and Ned H.C. Hwang**
* Mechanical Engineering Dept., Faculty of Engineering
Alexandria University
** Cardiovascular Flow Dyanmics Laboratory, Cullen College of
Engineering, University of Houston

ABSTRACT

The dynamic performance of three types of heart valve prostheses (HVP) are evaluated in a mock flow loop system. An elastic collapsible left ventricular chamber and an elastic aortic conduit are fabricated with geometrical similarity to a human left ventricular cavity and aorta during the systolic phase of the cardiac cycle. A typical left ventricular volume curve is used as the driving wave form to activate the ventricle to insure that the generated pressure and flow wave forms closely mimic the physiological ones. When the elastic conduit is replaced by a geometrically similar rigid glass conduit, changes in the dynamic performance of HVP are observed. The study emphasize the importance of the near field pulsatile flow pattern and the phase-by-phase motion of the boundaries involved on the dynamic behaviour of HVP.

INTRODUCTION

Heart valves are passive devices that open and close according to the pressure and flow conditions in the field. Up to the present time, most prosthetic heart valves were designed to operate under the pressure gradient in the conduit i.e valve opens when the pressure gradient (dp/dx) is in the direction of the flow, and closes when the gradient is reversed. Undoubtedly, this mechanism also describes the basic behavior of natural heart valves over a major portion of a cardiac cycle. Careful study of natural heart valves, however, showed that the operation of highly efficient natural valves are far more competent than that described above. First, physiological measurements show that the left ventricular pressure (P_v) exceeds the aortic pressure (P_a) during the first one-third of the ejection phase; and P_a exceeds P_v for the other two-thirds of the ejection phase Figure (1)[1]. Clearly, we can see that the operation of natural aortic valves is not completely governed by the pressure gradient. Secondly, several recent investigations indicated that the closing behavior of natural heart valves may also be affected by the near field flow patterns [2-9]. Several recently designed mechanical heart valve prostheses tested seem to indicate such characteristics. Early prosthetic heart valve studies emphasized the measurements of valve pressure drop, incompetency, and the results of accelerated fatigue tests. Much of these works were made primarily to compare the relative performance of the various types of valve prostheses in a particular in vitro testing flow loop . The general flow patterns, velocity profiles, and fluid shear stress distributions, were obtained by a few investigators using flow visualization techniques in this type of flow loop, or by computer simulations [9-18]. Many investigators indicated that these in vitro results may not be indicative of the vivo performance because of variations in geometry, driving wave forms, etc.. Bayat et al., [19] and Abdallah et al.,[20], examined just how much variation is possible between investigators using different testing rigs. The possible reasons for these variations were discussed. Recognizing the effect of near field pulsatile flow patterns on the closing behavior of valve prostheses, the current trend in the in vitro flow studies has shown, a return in emphasizing the basic dynamics of valve motions and the results that can be used to estimate the in-vivo performance of the prostheses.

Basic Considerations

Unlike the natural heart valve, an artificial replacement presents a new geometry to the cardiovascular system. In the vicinity of the replacement valve blood separates from the vessel boundaries and from

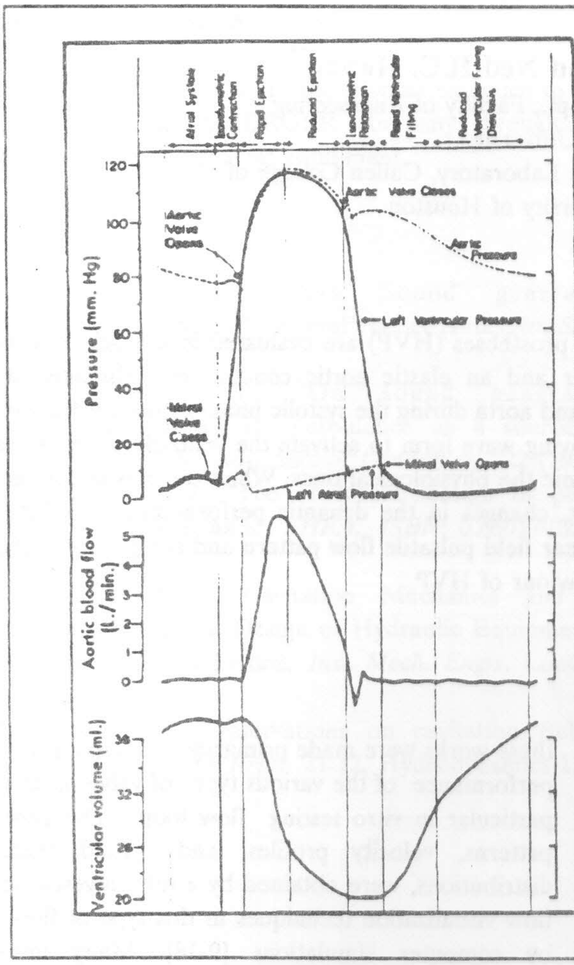

Figure 1. Pressure across the aortic valve (from Berne and Levy, Cardio-vascular physiology, 3rd Editioi., C.V. Mosby Co.)

the surface of the valve occluder. The patterns of flow separation at any particular phase in the cardiac cycle directly affect the pressure distribution around the occluder. The pressure distribution at the end of the ejection phase would certainly affect the closing behavior of the valve. The disturbance resulting from the flow separation could also generate abnormaly high shear stress and turbulence, as reported by Figliola and Mueller [21], Lu et al [22], Yoganathan and assocítates [23-25], and by Hwang et al. [4], [26] and [27]. In order to properly study the flow dynamics of prosthetic heart valves in an in vitro loop, careful considerations must first be given to the design of the mock system. This includes not only the geometrical similarly between the valve testing conduit and that of the corresponding natural cardiovascular system, but also the phase-by-phase motion of the conduit boundaries involved. The geometric similarly of the valve testing chambers provides the pattern of the basic fluid flow, while the motion similarily ensures adequate spacial and temporal distributions of force and acceleration. As far as the left heart is concerned, the ventricular contraction provides most of the energy to the fluid flow while the aorta and the atrium may be assumed passive.During the systolic phase, the ventricular muscle contracts with enormous power, which ejects the blood into systemic arteries through the aorta. This action may be accurately simulated by a collapsible ventricular-shapped chamber which empties according to a text-book type ventricular volume curve. The aortic wall motion is activated by the pressure generated in the ventricle. During the systolic ejection, a great portion of the pressure energy is stored in the aortic wall while it moves outward. The outward motion reaches a maximum distance at the peak of ventricular pressure, and smoothly returns to the diastolic position by gradually releasing its potential energy to the flow. The movement of the aortic wall is definitely affected by the wall elasticity, and the afterload against which the fluid is delivered .

The afterload in a natural cardiovascular system is the result of a rather delicate balance which involves complex interaction among nervous control, vasometer control, and local tissue metabolism. The afterload in an artificial flow loop, however, must be manually adjusted to achieve a set of desirable physiological pressure and flow waveforms.

In an isolated cat heart experiment, Elzingo and Westerhof [28] showed that the systemic afterload can be adequately simulated by a hydraulic model which consists of a characteristic resistance unit, a capacitance unit, and a peripheral resistance unit in series. With proper adjustment of the three, the physiological aortic pressure and flow waveforms can be achieved. The resistance units can be made with a few thousand tiny circular tubes. The small bores ensure laminar flow in the tubes throughout a cardiac cycle-an effect to keep the linear relationship between the pressure and the flow in the arterial system.

In a valve testing flow loop, an elastic aortic root was used to replace the natural ascending aorta. The dynamic performance of the prosthetic valves was found to be quite different from that in a rigid aortic conduit of the same geometry. The experimental methods and results of this study are presented as follows .

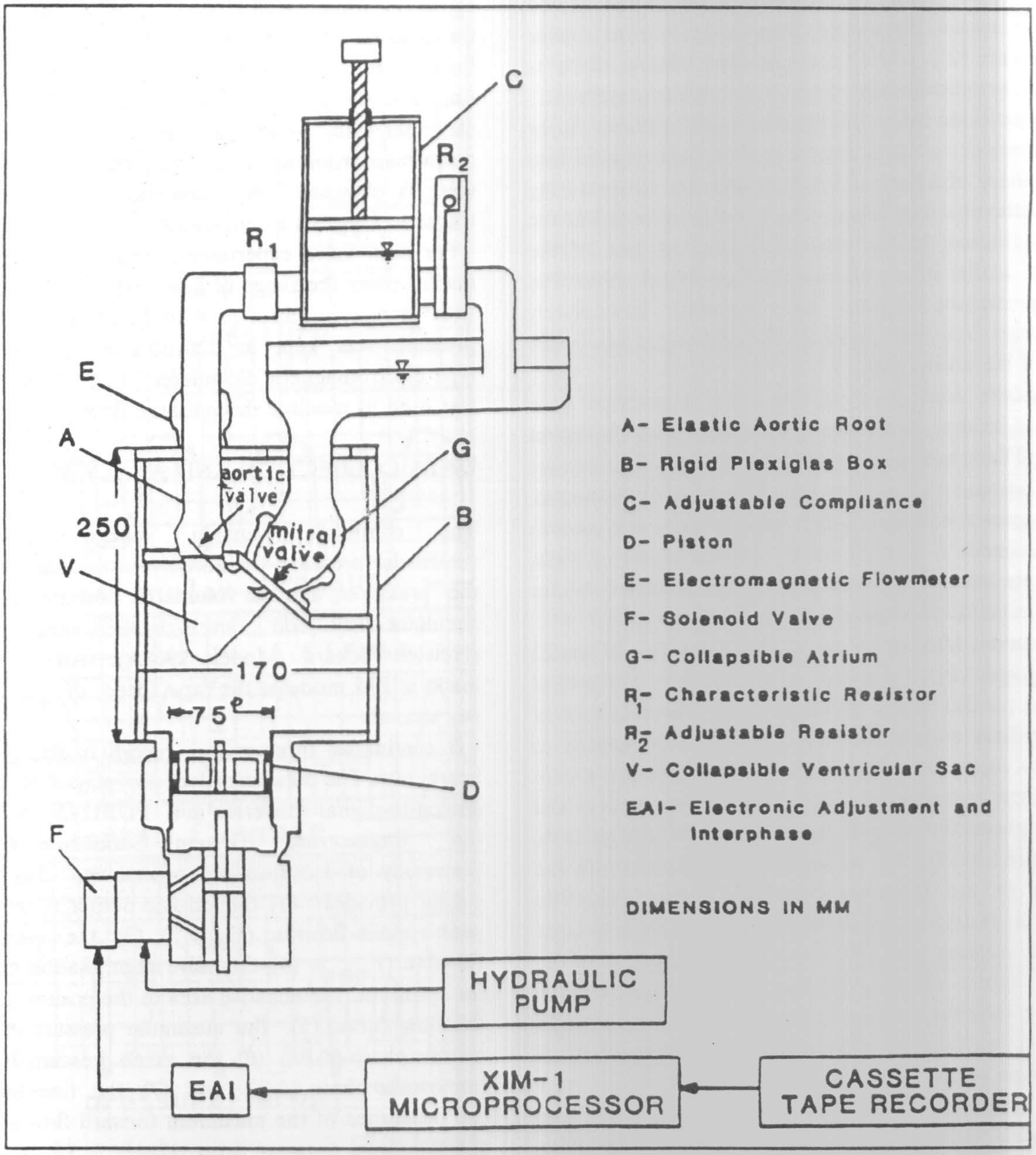

Figure 2. Schematics of the mock flow loop system.

EXPERIMENTAL METHODS

The pulsatile flow loop used in the experiments consists basically of three components: (a) the pulse generator, (b) the collapsible valve conduits, and (c) the afterload simulator. Figure (2) is a schematic showing the layout of the flow loop system .

The pulse generator includes a cassette tape recorder which plays back a pre-recorded waveform and feeds it into a KIM-1 microprocessor. The waveform is then digitized and used to control the movement of the piston through a solenoid valve. Electronic circuits are designed to provide the interface between the microprocessor and the solenoid valve. A hydraulic pump, which circulates a high pressure hydraulic fluid, provides the power to activate the piston.

The upper face of the piston contacts the liquid sealed in a rigid box (B. Figure (2)). The collapsible ventricular sac (V.Figure (2)), which contains the testing fluid, is dilated periodically according to the piston movement.

The ventricular sac is fabricated from a uniform layer (0.8 mm thick) of clear silicon rubber, and molded into the geometry of a human adult ventricular cavity during the systolic phase. A tilting-disc valve is mounted at the inlet of the sac in the same direction as that of the anterior leaflet in a natural human mitral valve. The valve prostheses studied are mounted immediately downstream from the outlet of the ventricular sac in the model of the aortic root.

The aortic conduits used in this experiment are fabricated to the same geometry suggested by Chong et al.[29]. The geometry includes three symmetric sinuses at the root into which the prosthetic valve is mounted. Two aortic conduits of different rigidities are used in the experiment : a rigid aortic conduit made of glass; and the elastic conduit fabrictated with a uniform 0.46 mm layer of clear silicon rubber.

The systemic afterload is simulated by a series of two 25 mm long resistors (R_1 and R_2). Each consists of several thousand parallel nickel capillary tubes. The diameter of the capillary tubes in each resistor is calculated to provide a range so that when the resistors are put in series, they represent the characteristic resistance and the peripheral resistance, respectively. An adjustable compliance chamber (c) is connected in series with the resistors to the flow as shown in Figure (2). The peripheral resistor is adjustable by positioning a circular bore in a rotating disc (P), which covers a portion of the capillary tubes. theoretically the arrangement of the capillarry tube resistors and compliance unit can provide an aortic waveform that closely mimics that of textbok-type aortic waveform.

Three prosthetic heart valves are used in the experiment. They are the St. Jude 27A, the Bjork-Shiley 27, and the Starr-Edwards Model 6120. Each valve is sequentially studied in the two aortic conduits installed in the flow loop.

A 36.7 % glycerin solution in distilled water is used as the blood-analog fluid. At 20^{o} C (room temperature) the solution has a density of 1.058 g/cm^3 and viscosity of 0.04 dyne sec./cm . In addition, the same experiment is performed using pure distilled water. The results from the two different testing fluids are also compared.

Pressure measurements are made through fluid-filled catheters, using Bell-Howell physiological pressure transducers. The ventricular pressure is measured at a flush mounted wall tap 15 mm upstream from the valve ring. The aortic pressure is measured using a 7F cardiac cathether with only side openings at four diameters downtream from the valve. The pressure drop across the valve is obtained by subtracting the simultaneous transducer signals electronically.

For each valve experiment, measurements are carried out to cover the range of heart rate from 70 to 120 bpm, with cardiac output of 4 to 11 liters/min. The aortic pressure was kept at 120/80 mm Hg. A flow-through type electromagnetic flowmeter probe (Zepeda EX 1.0) was used to monitor the pulsatile flow.

DATA COLLECTION AND ANALYSIS

The continuous analog voltage signals of the ventricular pressure, the aortic pressure, the difference of the pressures, and the volumetric flowrate, are recorded simultaneously on an 8-channel analog recorder (Hewlett-Packard Model 2968A). All recordings are made in FM mode at the tape speed of 7.5 inches per second.

A continuous three-minute record is obtained at each heart rate. The data recorded are played back into an analog-to-digital convertor in a PDP11/34 computer in the Engineering Systems Simulation Laboratory, University of Houston for processing. The computer output provides : (1) the cardiac output (CO), (2) the peak systolic flowrate (F_{max}), (3) the mean systolic flowrate (F_{mean}), (4) the valve regurgitation defined as the ratio of the negative area to the positive area under the flow curve, (5) the maximum pressure drop in the systolic phase ($\Delta\hat{P}$) (6) the mean pressure drop during the systolic phase ($\overline{\Delta P}$) , and (7) the time lag between the occurance of the maximum forward flow and that of the maximum pressure drop.

The zero-crossing of the pressure drop curve is used to provide a clock trigger which identifies the beginning of each pulse, and the zero-crossing of the flowrate curve (from forward flow to backward flow) is used to identify the end of the pulse.The values of ($\overline{\Delta P}$) and F_{mean} are calculated using the interval between these two zero-crossings.

In certain prosthetic valves, however, a period of negative pressure drop is clearly identified during the systolic phase (details see Discussion). In this case, the

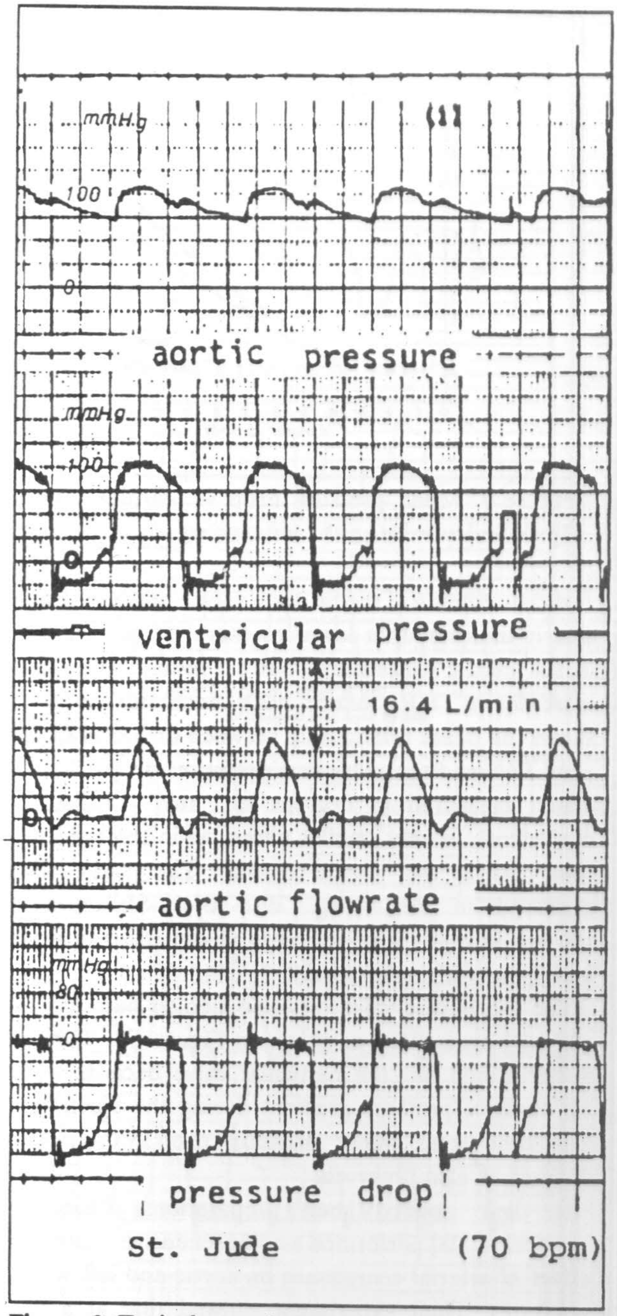

Figure 3. Typical wave forms generated by the system.

large negative area under the pressure drop curve (Figure (3), for example) is not counted for in the calculation of (ΔP). The very low or negative values of ($\overline{\Delta P}$) resulting from including the large negative area have little significance in valve dynamics.

For each valve experiment, at least two independent runs are made to cover the range of heart rates from 70 bpm to 120 bpm at increments of 10 bpm. For each heart rate, 35 cycles of continuous data are selected at random for the analysis.

Video recordings are made during the experiment by photographing the aortic root and the oscilloscope traces simultaneously on the same screen to observe the movement of the valve occluder. Although the visualization data are not exhibited directly in this paper, much of the real-time observations helped to interpret the data presented.

RESULTS AND DISCUSSIONS

A highly efficient prosthetic valve may experience a negative pressure gradient over a large portion of the ejection period similar to that of the natural heart valve as shown in Figure (1). In such case, a zero or negative mean pressure drop may be expected if average value is taken over the whole ejection period. A zero or negative mean pressure drop would have no physical meaning in the dynamic analysis of any heart valve. For the different types of prosthetic valves tested, the mean pressure drop cannot be defined in the same way over a given period (say, the period of ejection).

Figures (4, 5 and 6) show the measured maximum pressure drop across the Bjork-Shiley, and the Starr-Edwards Model 1620 valves, respectively. The cardiac output ranges from 3 to 11 liters per minute (l/min.) for heart rates ranging from 70 to 120 beats per minute (bpm) in each experiment.

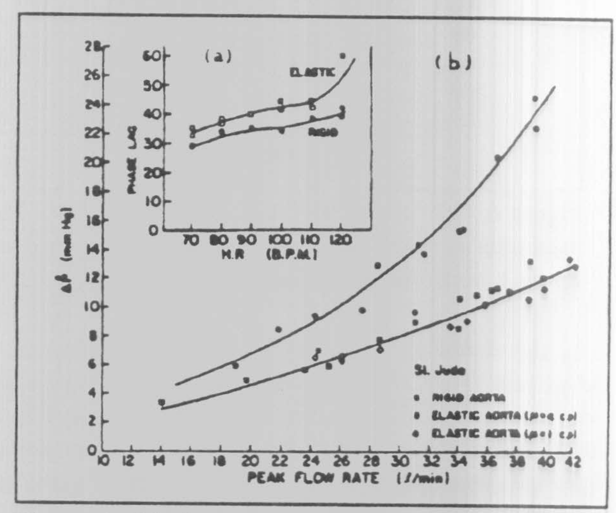

Figure 4. Maximum pressure drop vs. peak aortic flowrate: St. Jude valve.

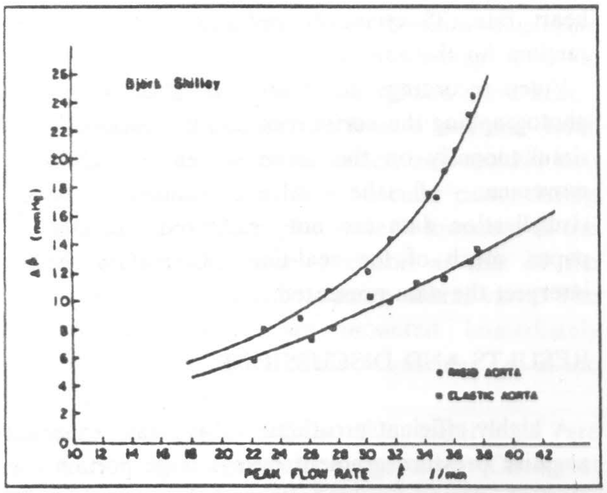

Figure 5. a) Phase lag between peak flow and maximum pressure drop. b) Maximum pressure drop vs. peak aortic flowrate "Bjork-Shiley".

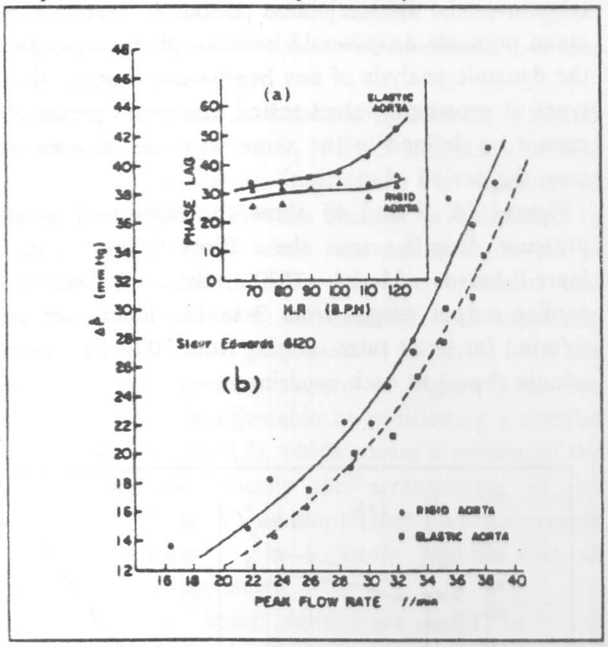

Figure 6. a) phase lag between peak flow and maximum pressure drop. b) Maximum pressure drop vs. peak aortic flowrate "Starr-Edwards".

In the abscissa, peak flow rate is used instead of the heart rate (HR) or cardiac output (CO). This is because the variations in CO were observed betweeen two different heart valves at the same HR; so were the corresponding flowrate waveforms. The peak flowrates were thus selected expecting that the maximum pressure drop occurs in a close association with the peak flow during systole.

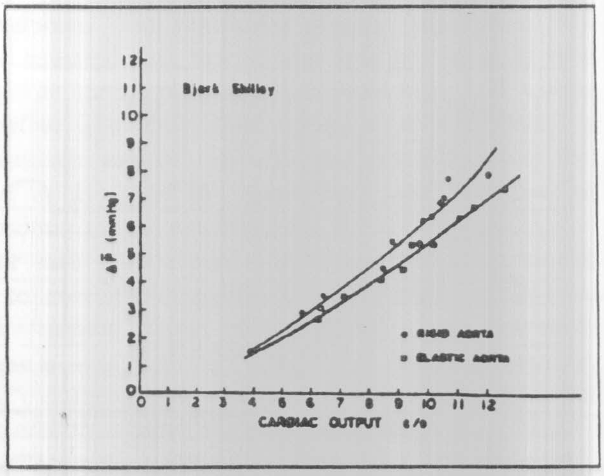

Figure 7. Mean pressure drop vs. cardiac output in Bjork-Shiley valve.

Measurements of the St. Jude valve shown in Figure (4) clearly indicates that lower pressure drops are obtained in the elastic (distensible) valve conduit. The difference increases as HR, hence CO, increases. Figure (4 a) shows the phase relationship between the occurance of the peak flow and the maximum pressure drop. The phase angle, measured in degrees in any cardiac cycle, consistently shows smaller phase lag in the rigid conduit than that in the elastic conduit. The same trend is observed in the tilting disc (Bjork-Shiley) valve and the ball-in-cage (Starr-Edwards Model 1620) in Figure (5) and Figure (6), respectively.

The tilting disc valve was also singled out for a plot of ($\overline{\Delta P}$) against CO as shown in Figure (7). This analysis is performed on the particular valve because it showed no negative pressure gradient during the ejection phase. The same trend, higher pressure drop in the rigid aortic conduit, is also observed.

In their isolated heart preparations, Elsinga and Westerhof [28] performed a comprehensive analysis on the effect of arterial compliance on aortic and left ventricular pressures. Their experiment showed that at a constant systemic resistance both the aortic pressure and the ventricular pressure increase as the conduit compliance decreases. In the meantime, the corresponding aortic flow decreases drastically. In order to maintain approximately the same cardiac output as we did in our experiment a larger pressure drop across the valve is needed.

In addition, the near field observations made through the video recording camera revealed certain fluttering

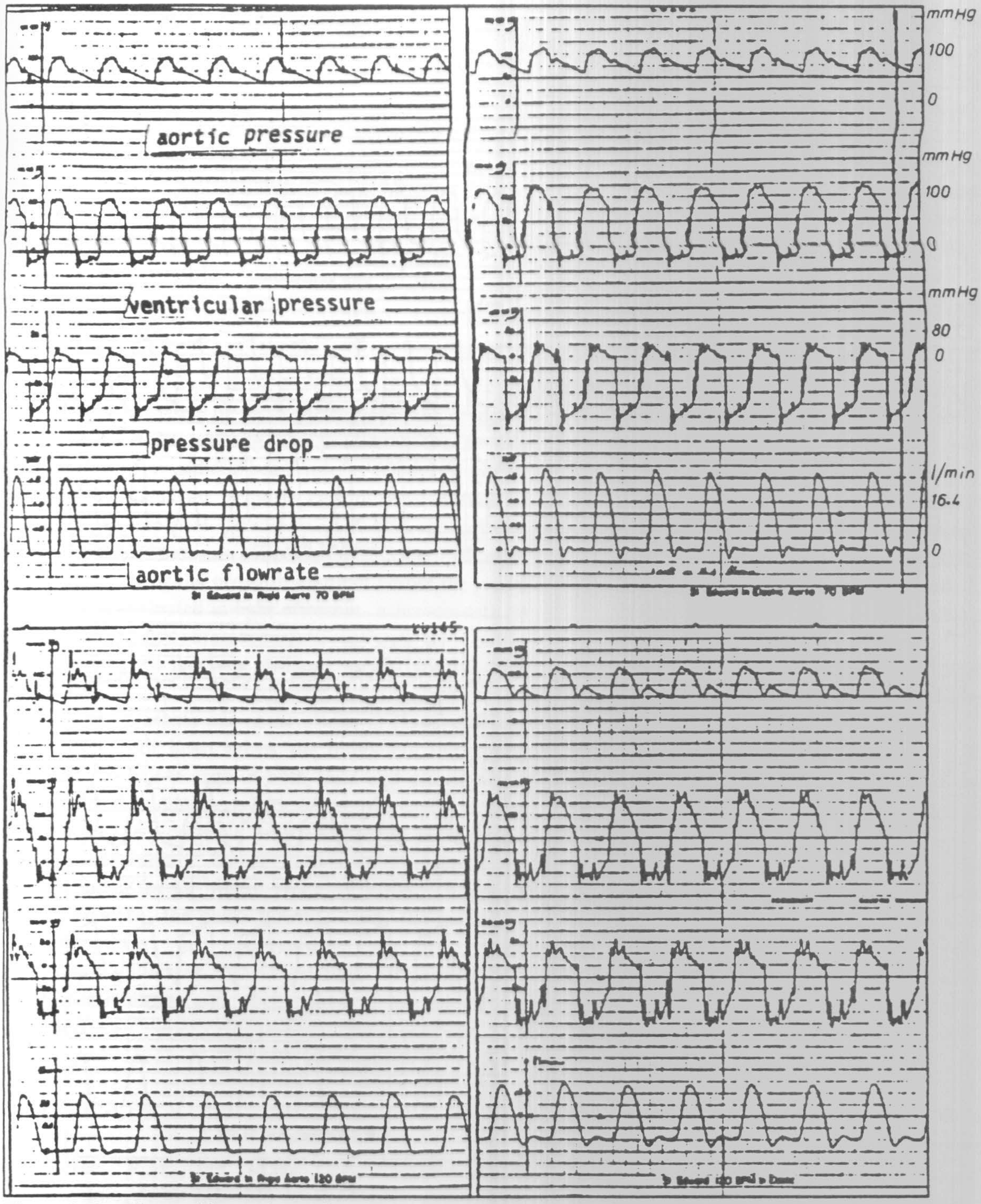

Figure 8. Oscillographic tracings of starr-edwards valve.

movement of the valve occluder during the ejection phase for each valve studied in the rigid valve conduit. This type of occluder fluttering movement can also be seen in the oscillograph tracings of both the ventricular and the aortic pressures, as shown in Figures (9 and 10). The fluttering motion prevents the occluder from achieving

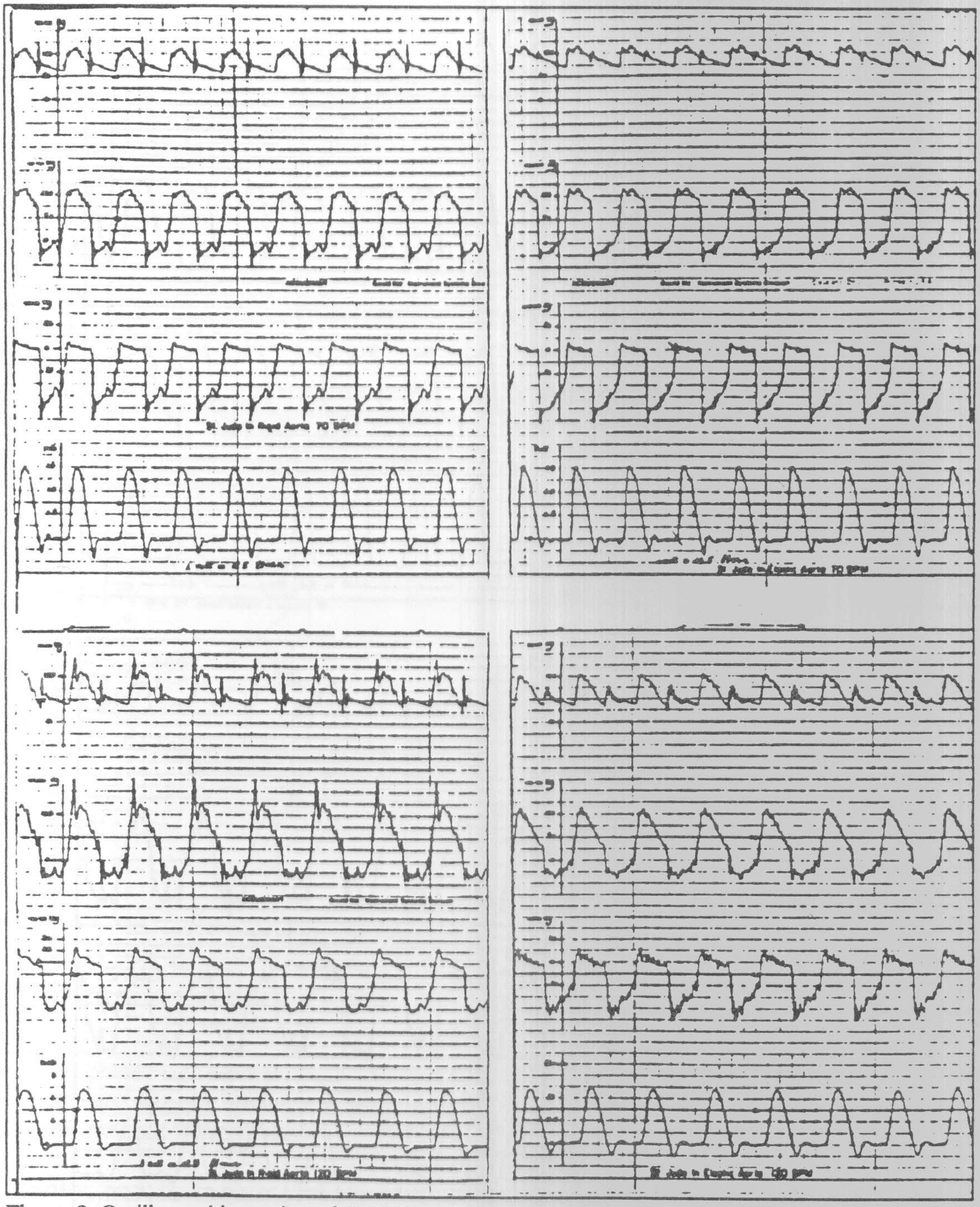

Figure 9. Oscillographic tracing of St. Jude valve.

a fully open position during the peak ejection. This may in part be responsible for the higher pressure drop observed.

The percentage of regurgitation, defined as the percentage ratio of the back flow to the forward flow in the cardiac cycle, is shown in Table 1. The measured back flow is significantly lower in the rigid conduit than that of the elastic conduit. We believe this is because the inward motion of the elastic aortic wall helps to push a certain amount of fluid back through the valve during the end of the systolic phase.

Table 1. Percentage Regurgitation

C/O	Rigid Conduit		Elastic Conduit	
l/min.	St.Jude	Star-Ed	St.Jude	Star-Ed
5	5.5	1.5	10.4	2.7
7	4.0	0.7	8.3	1.4
9	4.2	0.5	6.2	2.0
11	5.5	0.4	3.6	2.4

CONCLUSION

The new generation of mock flow loops are designed with emphasis on studying and predicting the in vivo performance of prosthetic heart valves. Model study of the dynamic behavior requires careful consideration of not only geometrical similarity, but also similarity in vessel boundary motions. The simple view of valve operation solely by the pressure gradient across the valve should be re-evaluated in light of recent achievements in more sophisticated measurements, as well as better understanding of the valve dynamic under pulsatile flow conditions.

This paper presents our observations of three different types of prothetic heart valves, the bi-leaflets, the tilting disc, and the ball-in-cage in both rigid and elastic aortic conduits of the same geometry under the same driving conditions. The pressure drop measured in the rigid coduit is consistently higher than that obtained in the elastic conduit for all three types of valves. The phase angles between the peak flowrate and the maximum pressure drop in each cardiac cycle are also significantly different. The possible reasons for these observations and their physical meanings are discussed.

REFERENCES

[1] Berne, R.M. and M.N. Levy, "Cardiovascular Physiology", *3rd Ed., C.V. Mosby Co., St. Louis, MO,* p. 91,1977.

[2] Bellhouse, B.J. and L. Talbot, "The fluid mechanics of aortic valve", *J. Fluid Mech.* 35:4, 721, 1969.

[3] Bellhouse, B.J., "*The fluid mechanics of heart valves. Cardiovascular Flow Dynamics*", (Ed., D.H. Bergel), Academic Press, London, 1972.

[4] Hwang, N.H.C., A.K.M.F. Hussain, P.W. Hui, T. Stripling, and D.W. Wieting, "Turbulent flow through a natural human mitral valve", *J. Biomech.*, 10:59, 1977.

[5] Knight, C.J., N. Nacleod, and D.E.M. Taylor, "Physical principles of the Edinburgh prosthetic heart valve", *Med. and Biol. Engr. and Comput.*, 15:264, 1977.

[6] Reul, H. and N. Talukder, "Heart valve mechanics", *Quantitative Cardiov. Studies* (Eds. N.H.C. Hwang, D.R. Gross, and D.J. Patel) Univ. Park Press, Baltimore, MD, 1979.

[7] Swanson, W.M., S.A. Ou, and R.E. Clark, " Vortex motion and induced pessure in a model of the aortic valve", *J.Biomech. Engr., Trans. ASME*, 100, 216, 1978.

[8] Van Steehoven, H.A. and M.E.H. van Dongen, "Model studies of the closing behavior of the aortic heart valve", *J. Fluid Mech.* , 90:21, 1979.

[9] Weiting, D.W., N.H.C. Hwang and J.H. Kennedy, "Fluid mechanics of the human mitral valve", *AIAA Paper* #71:102, 1971.

[10] Duff, W.R., "Fluid dynamics of prosthetic cardiac valves", *Ph.D.Thesis*, Purdue University. Lafayette, Ind, 1969.

[11] Hung, T.K. and G.B. Schluessler, "Computational analysis as an aid to the design of heart valves", *Adv. Bioengr. Chem. Eng. Progr. Symp.* Ser. 67 (114) ;8, 1971.

[12] Kaster, R.L., Lillehei, C.W. and Starek, P.J.K., "The lillehei-Kaster pivoting-disc aortic prosthesis and a comparative study of its Pulsatile flow characteristics with four other prostheses", *Trans. Amer. Soc. Artif. Intern Organs* 16, 233,1970.

[13] Reul, H., P.C. Lu , J.D.A. Van Hoften, and N.H.C. Hwang, "Flow patterns of artificial heart valves analyzed by a new laser technique", *Proc., ACEMB*, 19:413 , 1977.

[14] Roschke, E.J. and E.C. Harrison, "Fluid Shear stress in prosthetic heart valves", *J. Biomech.*, 10:299, 1977.

[15] Swanson, W.M., " Observations of flow dynamics of caged-ball valves", *Proc., ACEMB*, 16:113, 1972.

[16] Weiting, D.W, "Dynamics flow characteristics of heart valves", *Ph.D. Thesis*, Univ. of Texas, Austin, TX , 1969.

[17] Wright, J.T.M. and L.J. Temple, "A FLow visualization study of prosthetic aortic and mitral