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A Method for Determination of Pressure Drop in Caged-Ball and Caged-Disc Prosthetic Cardiac Valves

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Abstract. This paper describes a simple theoretical and experimental method to determine the pressure drop across mitral caged-ball and caged-disc prosthetic heart valves under steady state flow conditions. The overall pressure drop across any mechanical artificial heart valve is affected not only by its orifice, but also by the poppet design and the entire superstructure of the valve. In this study, the total pressure drop across the prosthesis is considered to be equal to the sum of the pressure drops due to the valve occluder and the orifice. These two pressure drops are analysed separately and then superimposed to obtain the total pressure drop. Experimental total pressure drops at different volume flow rates were obtained using an appropriate pressure transducer. The valves were installed inside a test chamber which enables pressure drop measurements under steady state flow conditions using a blood analog fluid. The theoretical and experimental results of pressure drop are in good agreement, particularly in the physiological range of interest, for both valve types. Also, it was found, as anticipated, that the pressure drop through the tested prosthetic heart valves is mainly due to the orifice flow.

1. Introduction

In 1960, cardiac operations of valve replacement in human became generally accepted. Thenceforth, many prosthetic valve design have been introduced in an effort to improve their performance. Some designs have evolved and can still be found today. Others have been abandoned. The present prostheses are neither ideal prosthetic systems nor do they perform as well as the natural healthy biological valves. The perfect or universal prosthetic valve remains elusive. From an engineering point of view, the present mechanical prosthetic heart valves can be classified into four main groups as follows [1,2]:

- *Group (A)* : the caged-ball prostheses
- *Group (B)* : the caged-disc (central-disc) prostheses
- *Group (C)* : the unileaflet tilting disc prostheses
- *Group (D)* : the bileaflet tilting disc prostheses

Generally speaking, Group (A), then Group (B) appeared first in the biomedical literature, followed by Group (C), and Group (D) was introduced relatively recently. Over the last three decades, the caged-ball and caged-disc prostheses, together, are the largest category of artificial valves being implanted and used for cardiac valve replacement. Therefore, this study concerns the experimental and theoretical analysis for valves of Group (A) and Group (B) type.

Under normal human conditions, the systolic period is about 30% to 40% of the heart beat period, *i.e.* the diastolic period is about 60% to 70% of the total cardiac cycle. Consequently, the atrio-ventricular valves open for a good fraction of the total cardiac cycle. This time is necessary for filling the ventricles with blood which will be pumped by ventricular contraction during the systole. This explains the importance of pressure loss evaluation through the mitral valve during diastole period. One can assume that steady state occurs when the mitral valve is fully open and velocity is maximal [3,4,5]. The steady state flow tests are essential to predict certain flow characteristics and pressure gradient loss, before conducting more complicated, expensive, and difficult-to-interpret pulsatile flow tests.

Computational numerical analysis techniques (see, for example [6]) and sophisticated experimental mock circuit studies (see, for example [7]) have been used to assess the flow characteristics and pressure gradient loss in the vicinity of prosthetic heart valves. Both methods are laborious, use expensive equipment, and are time consuming. Therefore, this work represents a simple method for the determination of pressure drop across caged-ball and caged-disc prosthetic heart valves under steady state flow conditions. This pressure drop loss is an important factor among many other parameters affecting the overall performance of prosthetic heart valve.

Obviously, the pressure drop through an ideal prosthetic heart valve should be a minimum. In this study, two types of commonly used mitral prostheses are experimentally and theoretically studied for pressure drop under steady state flow conditions. From a theoretical point of view, the total pressure drop across the prosthesis is considered to be equal to the sum of the pressure drops due to the valve occluder and the orifice. Theoretical results for total pressure drop at different volume flow rates are then compared with the experimental findings.

2. Experimental Studies

2.1. Steady State Flow Circuit

Figure 1 illustrates the steady flow circuit and its components. The flow loop consists of: **a)** a centrifugal pump of 1/3 hp with a maximum flow rate of 45 liters per minute; **b)** two rotameters connected in parallel, one to measure high flow rates and the second to measure small flow rates. The final accurate flow adjustment is done by manipulating the three valves A, B and bypass valve C as shown in Figure 1; **c)** a flow rectifier for control of vortices and other flow perturbing effects at the test chamber inlet; **d)** a mitral test chamber for the prosthesis; and **e)** a reservoir con-

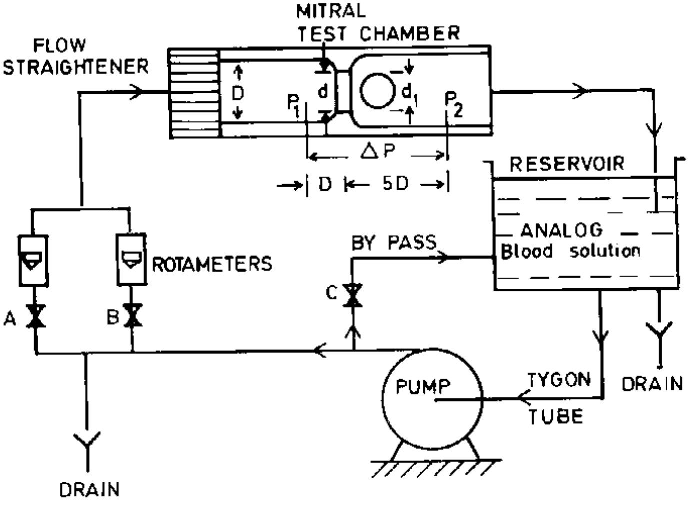


Fig. 1. The steady state flow circuit

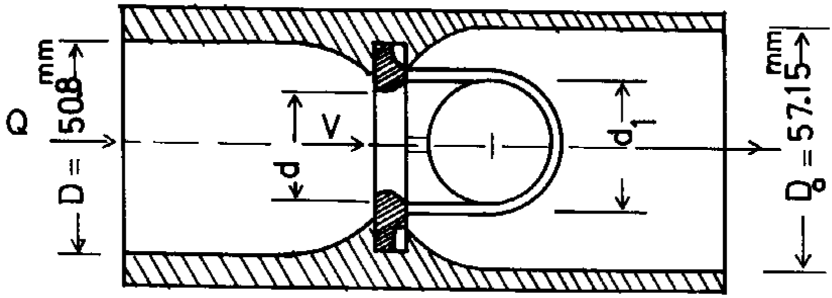
taining the blood analog solution. All circuit components are connected with tygon tubes.

2.2. Test Chamber

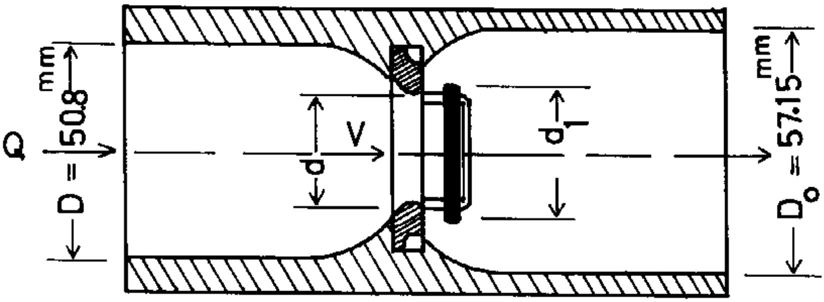
The prosthetic cardiac valves were tested under steady state flow conditions in a mitral test chamber which was first introduced by Wieting [8] and accepted in the biomedical engineering literature for pressure drop assessment and observation of other flow characteristics. Different valve sizes could be accommodated in the chamber by using special adaptors. The test chamber is made of a transparent material (plexiglass) in order to visualize the fluid flow in the vicinity of the valve.

2.3. Valves Tested

Figure 2 illustrates a caged-ball valve, and a caged-disc valve, installed inside the mitral test chamber. The Starr-Edwards valve model 6120 is taken as a typical example of a caged-ball design. The Beall valve model 106 is taken as a typical example of a caged-disc design. Both valves, although being relatively old models (introduced in 1966 and 1974 respectively), are still clinically used successfully [2,9] and are easily available. The specifications of each prosthesis (model and size) are given in the appendix. More recent models of caged-ball valve design (group A) and caged-disc valve design (group B) have been periodically introduced in the literature, but their basic design has remained virtually unchanged (an occluder, ball or disc, a cage and an orifice). Many of these recent models were found to be less common, being dis-



(a) Starr-Edwards 6120



(b) Beall 106

Fig. 2. Valves tested

continued or abandoned due to their serious postoperative complications [2,9]. Therefore, the above selected models may be considered as good examples for this study. In fact, a new valve design, although carefully developed, does not automatically mean that it is a better one than older models. The only judge of its superiority is the clinical implantation and the consecutive follow-up evaluation over the years in patients.

2.4. Test Fluid

The fluid used during the experimental pressure drop evaluation, across the above prostheses, at different steady volume flow rates (from 5 lit/min to 25 lit/min), was an aqueous blood analog solution containing 36.7 percent glycerol by volume. The solution has a viscosity of 3.5 centipoise and a specific gravity of 1.107 at 25°C. Human blood viscosity is approximately 3.3 centipoise at 37°C and specific gravity about 1.058.

2.5. Pressure Tap Location

Pressure tap location is critical and can affect pressure drop results. Pressure should be measured in regions where flow transition disturbances are negligible, and at locations where the pressure is essentially constant across the flow area. Pressure differences should be measured between two points of equal flow area to avoid a pressure drop correction, due to the different velocities in the enlargement or contraction sections of the test chamber. Relatively large disturbances are produced by primary orifice flow separation and by cages and occluders with central occluding valves, and by jet boundary separation with central flow valves. The effect of these disturbances on the uniformity of pressure across the flow area is usually negligible about 5 diameters downstream from the valve. This dimension is nominally about 3/4 of the channel diameter. A downstream pressure tap location of 4 to 5 channel diameters from the valve is then considered optimum. This downstream position corresponds to the point of maximum pressure recovery from a wake or jet disturbance [4].

The upstream pressure field is only slightly disturbed by the open valve. The natural flow enters the valve through a fairly well shaped inlet tract nozzle geometry. The flow boundary layer is thin and the flow is fairly uniform. An upstream pressure tap location is therefore situated nominally 1 to 2 diameter upstream from the valve.

2.6. Pressure Drop Measurements

It is desirable to measure the pressure difference directly rather than as the difference of two separately measured pressures. An appropriate pressure transducer will give a more satisfactory resolution and calibration accuracy. A Celesco P-7D differential (variable reluctance) pressure transducer (Celesco Transducer Products Inc., Canogo Park, California, USA) was used for that purpose. The correction for diameter enlargement of the test chamber channel from 50.8 mm to 57.15 mm should not significantly affect the pressure drop results. With such relatively large diameters, it would be less than 3% at a flow rate of 15 liters per minute [10].

3. Theoretical Considerations

3.1. General Introduction

The overall pressure drop across a given prosthetic cardiac valve is affected not only by its orifice, but also by the poppet shape and design and the entire superstructure of the valve. Thus,

$$\Delta p = \Delta p_1 + \Delta p_2 + \Delta p_3 + \Delta p_4 \tag{1}$$

where,

Δp = total pressure drop across the prosthesis.

Δp_1 = pressure drop due to the orifice flow.

Δp_2 = pressure drop due to the occluder (ball or circular disc).

Δp_3 = pressure drop due to the superstructure of the cage.

Δp_4 = pressure drop due to the internal wall friction of the mitral test chamber.

Assuming that the pressure drop across prosthetic cardiac valves is essentially due to their orifice and partially due to the occluder, Δp_3 and Δp_4 may be neglected (see, for example [11]).

Equation (1) becomes;

$$\Delta p = \Delta p_1 + \Delta p_2 \tag{2}$$

3.2. Pressure Drop due to the Orifice

For steady incompressible flow through an orifice of cross section area A, the pressure drop across it can be related to the volumetric flow area Q by the equation:

$$Q = C_d . A . \sqrt{\frac{2 . \Delta p_1}{\rho}} \tag{3}$$

where,

C_d = the orifice discharge coefficient.

ρ = density of the analog blood solution.

Q = volume flow rate = V.A

A = cross-sectional area of the orifice $= \pi d^2/4$

V = mean orifice velocity.

d = orifice diameter

Rearranging equation (3) gives,

$$\Delta p_1 = \frac{\rho . Q^2}{2 \ A^2} \cdot \left[\frac{1}{C_d^2} \right] \tag{4}$$

For simplification, the flow through the prosthetic heart valve orifice is assumed to be approximately similar to the flow through a sharp-edged pipe orifice. Therefore, the values of the orifice discharged coefficient, for the different sizes of tested prosthetic heart valves, can be determined from Fig. 3. The orifice discharge coeffi-

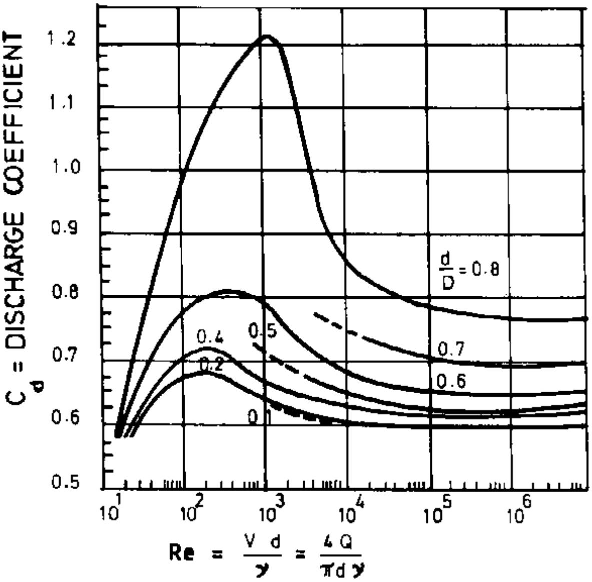


Fig. 3. Discharge coefficient versus Reynolds number (12)

cient is given [12] as a function of the Reynolds number Re = V.d/v = 4Q/ (π.d.v) and the radio d/D, where v is the kinematic viscosity of the blood analog solution and D the inlet diameter of the mitral test chamber.

3.3. Pressure Drop due to the Occluder

For group (A) the occluder is a ball (sphere). For group (B) the occluder is a circular disc. The valve poppet of each group is assumed to be situated in a free stream velocity equal to V_{vc}.

Thus,

$$C_D = \frac{\Delta p_2}{\frac{1}{2} \cdot \rho \cdot V_{vc}^2} \qquad (5)$$

where,

V_{vc} = fluid velocity of the orifice vena contracta = Q/A_j
A_j = contracted area of the vena contracta ≈ 0.65 A

C_D = the occluder drag coefficient (for the ball or the disc).

Rearranging equation (5) gives,

$$\Delta p_2 = \frac{1}{2} \cdot \rho \cdot V_{vc}^2 \cdot C_D = \frac{1}{2} \cdot \rho \frac{Q^2}{A_j^2} \cdot C_D$$

$$\Delta p_2 = \frac{\rho \, . \, Q^2}{2 \; A^2} \cdot \left[2.366 \; C_D \right] \tag{6}$$

The values of the occluder drag coefficient for valves of group (A) and for valves of group (B) can be determined from Figure 4. The occluder drag coefficient is given [12] as a function of the Reynolds number $Re_1 = V_{vc} \, . \, d_1/\nu$ and the shape of the occluder, where d_1 is the diameter of the valve occluder.

3.4. Total Pressure Drop

Adding equation (4) and (6); gives the total pressure drop,

$$\Delta p = \frac{\rho \, . \, Q^2}{2 \; A^2} \cdot \left[\frac{1}{C_d^2} + 2.366 \; C_D \right] \tag{7}$$

The orifice contribution as a fraction of total pressure drop is:

$$\Delta p_1/\Delta p = \Delta p_1/(\Delta p_1 + \Delta p_2) = 1/(1+2.366 \, C_D . C_d^2) \tag{8}$$

4. Results

Figure 5 illustrates the experimental and theoretical results of the total pressure drop across group (A) valves versus the volumetric flow rate. Similarly, Fig. 6 illustrates the experimental and theoretical results of the total pressure drop across group (B) valves versus the volumetric flow rate. The numerical values of Re and Re_1 were found to be in the range from 1.38×10^3 to about 1.15×10^4 (equivalent to flow rates of 5 lit/min to 25 lit/min). In such a range of Reynolds numbers, the occluder drag coefficients (C_D) for the ball and the disc remains more or less a constant value as deduced from Fig. 4. The occluder drag coefficient for the ball is 0.4. The occluder drag coefficient for the disc is about 1.12. The pressure drop across the valves of group (A) or group (B) is mainly due to the orifice flow. Equation (8) gives the percentage orifice contribution to the total pressure drop. For group (A), the caged-ball type, the pressure drop due to orifice flow is about 85% of the total pressure drop. For group (B), the caged-disc type, the pressure drop due to orifice flow is about 70% of the total pressure drop.

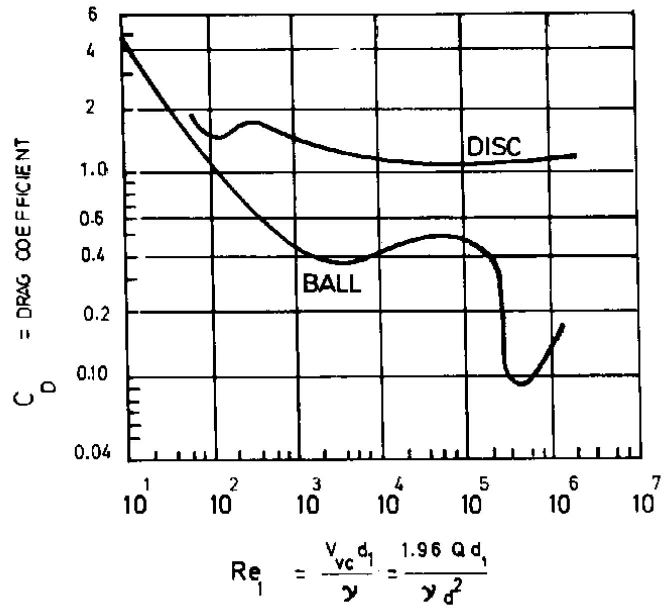


Fig. 4. Drag coefficient versus Reynolds number (12).

5. Conclusion

The theoretical and experimental results of pressure drop, through caged-ball valves and caged-disc valves, are in good agreement, particularly in the physiological range of interest. The pressure drop across the prosthesis is essentially due to the orifice flow, and only partially due to the occluder. This approach can be applied to determine and evaluate the pressure drop across many types of artificial cardiac valves which belong to the caged-ball or caged-disc valve design. The correlation between this analytical method and the experimental findings is excellent for large diameters orifices.

Finally, it should be noted that in the case of two mitral prosthetic heart valves, one a caged-ball type and the other a disc-caged type - and both having the same orifice diameter - the total pressure drop due to the disc type, as expected, is higher than the total pressure drop due to the ball type. This is shown in Fig. 5 and 6 for valves of orifice diameter equal to 19 mm. However, an advantage of the disc-type over the ball-type is the lower volume occupied by its valve structure inside the ventricle.

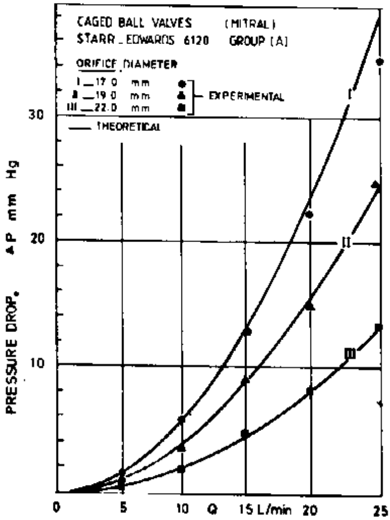


Fig. 5. Pressure drop at different flow rates (caged ball valves)

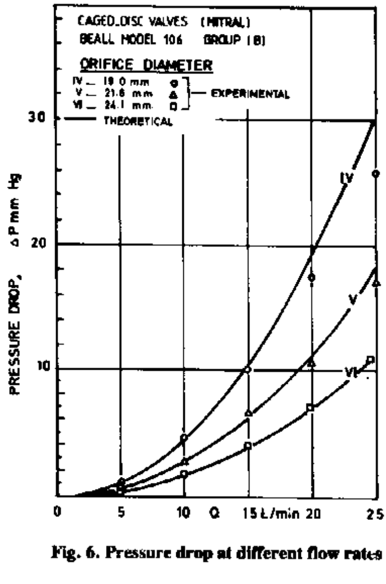


Fig. 6. Pressure drop at different flow rates (caged disc valves)

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Appendix

1- *Caged-ball Design*

Group (A) - Starr-Edwards Model 6120 (Introduced in 1966 to present)

Size	Annulus diam. (mm)	Orifice diam. d (mm)	Orifice area (cm^2)
26 M	26	17	2.269
30 M	30	19	2.835
34 M	34	22	3.801

2- *Caged-disc Design*

Group (B) - Beall Model 106 (introduced in 1974 to present)

Size	Mounting diam. (mm)	Orifice diam. d (mm)	Orifice area (cm^2)
small	30.5	19.0	2.835
medium	33.0	21.6	3.664
large	35.5	24.1	4.561

طريقة لتعيين فقدان الضغط عند مرور الدم خلال صمامات القلب الصناعية

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ملخص البحث: يقدم هذا البحث طريقة نظرية ومعملية بسيطة لتعيين فقدان الضغط عند مرور الدم خلال صمامات القلب الميترالية الصناعية تحت شروط السريان الثابت وتم تطبيق هذه الطريقة على نوعين من الصمامات الصناعية الميكانيكية الأكثر انتشارًا وهما الصمامات ذات الكرة والصمامات ذات القرص المركزي.

من المعروف أن فقدان الضغط خلال صمامات القلب الصناعية لا يتأثر فقط بفتحة الصمام ولكن أيضا بشكل محبس الصمام والشكل الهندسي العام الخارجي للصمام. في هذا البحث أعتبر فقدان الضغط الكلي خلال صمام القلب الصناعي يساوي مجموع عاملين أساسيين لفقدان الضغط، العامل الأول هو فقدان الضغط نتيجة فتحة الصمام، والعامل الثاني نتيجة مرور الدم حول محبس الصمام. وعلى ذلك تم تحليل كل عامل على حدة ثم تم تجميع العاملين معا للحصول على فقدان الضغط الكلي النظري.

ولقد تم قياس الضغط الكلي المعملي في دائرة مناسبة للسريان الثابت عند معدلات مختلفة للتدفق باستخدام جهاز ملائم لقياس فرق الضغط مع تثبيت الصمامات المستخدمة بإحكام داخل حجرة ميترالية صممت خصيصا لمثل هذا النوع من الاختبارات تحت شروط السريان الثابت، وقد استخدم في التجارب سائل مماثل للدم مكون من خليط من الماء المقطر والجلسرين له نفس لزوجة الدم.

ولقد وجد أن النتائج المعملية والنتائج النظرية لفقدان الضغط الكلي لجميع أنواع الصمامات التي تم دراستها في توافق جيد خاصة عند معدلات التدفق الفسيولوجية. كذلك وجد أن فقدان الضغط خلال صمامات القلب الصناعية يرجع أساسا للعامل الأول أي نتيجة لمرور الدم من خلال فتحة الصمام.