

TECHNICAL NOTE

DESIGN OF A SYSTEM FOR THE ACCELERATED LOADING OF HEART VALVE PROSTHESES

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Abstract—In this paper an equipment is described for the loading of heart valve prostheses under physiological pressure conditions at a frequency of 10 Hz. The system consists essentially of two reservoirs between which a housing is mounted for holding the valve prosthesis. The reservoirs are partly filled with liquid. The physiological pressure variation across the valve is obtained by pressure control within the two reservoirs. A phase difference between the pressures in the two reservoirs compensates for the mass-inertia effects which normally occur at these high frequencies. The system without a valve has been analysed on the basis of simplified relationships between pressure and flow. The predicted values for the phase difference between the flow and the pressure curves within the valve housing, have been verified experimentally for various values of phase difference and amplitude ratios of the pressure variations within the reservoirs. The agreement between theory and experiment is fair. For the system with a valve the experimentally observed patterns closely resemble the theoretically predicted ones. From experiments with a Björk-Shiley (21ABP) and a Hancock (242-A21) valve prosthesis it is concluded that the valves open and close completely and that the pressure and flow patterns around the valves mimic the essential features of the *in-vivo* signals.

INTRODUCTION

The aim of our research project is to lay down specifications for the construction of heart valve prostheses. In order to obtain basic information about useful designs and about the characteristics of the materials used, the natural aortic valve (Sauren *et al.*, 1980, 1983; van Steenhoven *et al.*, 1979, 1981, 1982b) as well as prosthetic valves were evaluated (Rousseau *et al.*, 1983; van Steenhoven *et al.*, 1982a). For a future evaluation of prototypes of new valve design, a system is required which opens and closes heart valve prostheses under physiological conditions, in an accelerated way up to a frequency of 10 Hz. As the pressures in the vicinity of the valve are the parameters responsible for the valve loading, the pressure ahead of and behind the valve must resemble the physiological signals, especially with regard to shape and magnitude. Furthermore, a small phase difference between the pressure and flow curves is required to ensure valve opening and closing at the correct moments in the simulated cardiac cycle. The frequency of 10 Hz is necessary for shortening the time of the experiment.

Several accelerated fatigue systems are described in the literature and can be put into two categories. The first group consists of flow-controlled systems. The pressure across the valve will be determined by the inertia of mass and the presence of a fluid resistance in the system (Almond *et al.*, 1974; Clark *et al.*, 1978; Häußinger and Reul, 1981; Haworth *et al.*, 1976; Martin *et al.*, 1980; Quinton *et al.*, 1961 and Swanson and Clark, 1978). The second group consists of pressure-controlled systems. The flow curve is the result of a variable flow resistance in the system (Koorajian *et al.*, 1969 and Steinmetz *et al.*, 1964). However, from the given experimental results, it was concluded that none of the systems described delivers physiological curves for flow and pressure at an acceptably low phase difference. Therefore, the aim was to

design a new system in which the pressure patterns resemble the physiological ones and the phase difference between the pressure and the flow curve is minimized.

DESCRIPTION OF THE SYSTEM

The main problem in designing an accelerated testing device is that in comparison with the physiological condition the mass inertia effects will play an important role. The phase difference between pressure and flow, which is quite small in the physiological condition, will increase with rising frequency of opening and closing. The main requirements for the design of such a system are that, (1) the pressure patterns ahead of and behind the valve look physiological (2) the phase difference between the flow and the pressure curve is minimized and (3) the valve must open and close completely. To realize this, a system has been designed which basically consists of two perspex reservoirs between which a transparent lucite housing is mounted (Fig. 1a). The length of the channel between the reservoirs is 80 mm and it has a bore of 16 mm. Different types of valves can be mounted in the valve housing. Both reservoirs are partly filled with liquid (e.g. water, saline solution, glutaraldehyde solution). The physiological conditions near the valve are obtained by varying the pressure in the volumes of air in the reservoirs. To reduce the phase difference between the flow and pressure curves, the pressure variations in the volumes of air in reservoirs 1 and 2 are obtained with the help of the two control mechanisms CM_1 and CM_2 . They consist (see Fig. 1b) of a rotating disc with eight circular gaps, which provide a time-varying resistance to the air flow. Both mechanisms are driven simultaneously by the same motor using a toothed driving belt. With those mechanisms it is possible to introduce pressure variations in the reservoirs at the same frequency and an adjustable phase difference. This makes it possible to minimize the phase difference between

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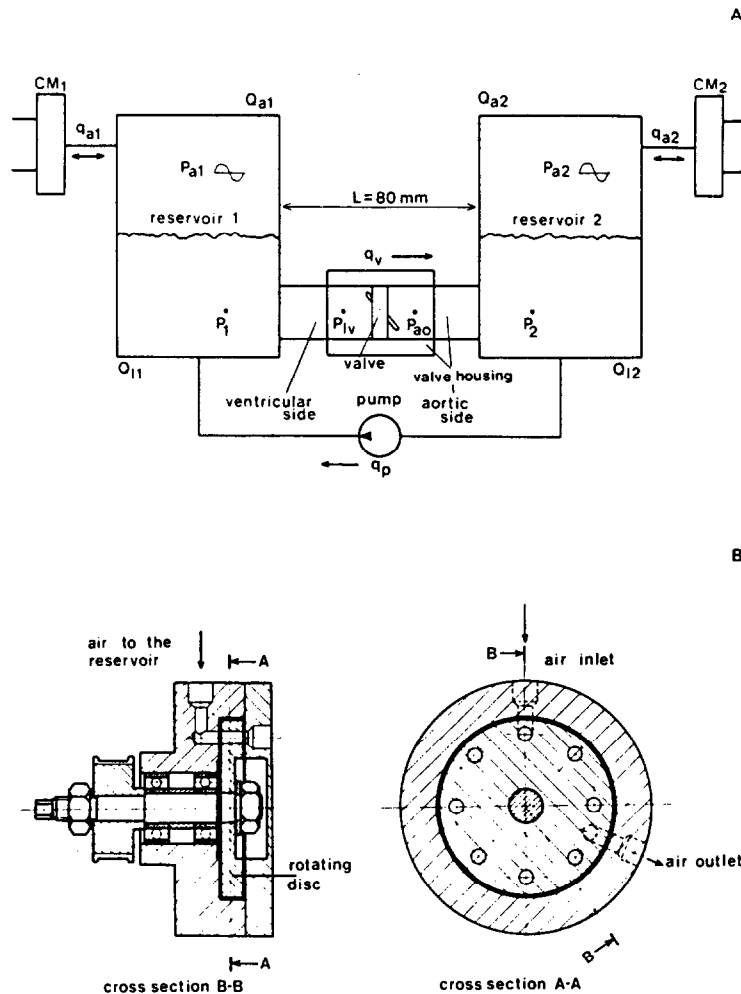


Fig. 1. (a) Schematic outline of the system. The following symbols are used: $CM_{1,2}$: control mechanisms; $q_{a1,2}$: air flow through the control mechanism; q_p : flow through the pump; $Q_{a1,2}$: air volume; $Q_{l1,2}$: liquid volume; $p_{a1,2}$: air pressure; $p_{l1,2}$: liquid pressure; p_{lv}, p_{av} : pressure ahead of and behind the valve; L : length of the channel. (b) Schematic outline of control mechanisms CM_1 and CM_2 .

flow through the valve and pressure over the valve. Variation of the phase difference between p_1 and p_2 is achieved by turning the rotating disc of one control mechanism a little with respect to the rotating disc of the other mechanism. To keep the liquid levels in the reservoirs constant, a pump with a constant flow is required. The whole system is of simple design and has a small liquid filling to ensure long-term functioning.

THEORETICAL ANALYSIS OF THE SYSTEM

Physical model for the system without a valve

To deduce a simple physical model of the system, a lumped circuit approach is used, neglecting both heat conduction in the air reservoir and the effect of viscosity on the channel flow. Furthermore, local flow phenomena at the connection of reservoir and channel will not be accounted for. The system equations can then easily be derived from left to right in Fig. 1a. First let us consider the system without valve. We assume the air to behave as an ideal gas and that isentropic compression and expansion occurs. Furthermore, the relative pressure variations in the air volume of reservoir 1 (p_{a1}) and

the air volume of reservoir 1 (Q_{a1}) as well as the flow ratio q_v/q_{a1} (which is the ratio of the flow through the channel and the flow of air through the control mechanism 1) are considered to be small. Then the gas law and the mass conservation law yield the following relation between the pressure in the air volume of reservoir 1 (p_{a1}) and the flow through the control mechanism 1 (q_{a1}), that is

$$\frac{dp_{a1}}{dt} = \frac{1}{C_1} q_{a1} \quad (1a)$$

$$\text{with} \quad C_1 = Q_{a1}/\gamma \bar{p}_{a1} \quad (1b)$$

where Q_{a1} = air volume in reservoir 1

γ = the specific heat ratio = 1.4

$\bar{p}_{a1}$ = air pressure averaged over one period.

It is assumed that the control mechanism 1 realizes a flow q_{a1} of the form

$$q_{a1} = \hat{q}_{a1} \cdot \sin(\omega t) \quad (2)$$

with $\hat{q}_{a1}$ as the amplitude of q_{a1} . Furthermore, it is assumed that:

(1) the pressure is continuous at the interface between water and air,

(2) the cross sectional area of the reservoir is large in comparison with that of the channel,
 (3) the liquid height in the reservoir is small, and
 (4) the flow velocities in the reservoir are negligible.
 Then we have

$$p_1(t) = p_{a1}(t). \quad (3)$$

Here $p_1(t)$ stands for the pressure in the liquid in reservoir 1 and $p_{a1}(t)$ for the pressure in the air volume in the same reservoir. With the assumptions that the velocity profile in the channel is flat, the valve is positioned in the middle of the channel and the non-linear velocity terms may be neglected, the integrated equation of motion yields

$$p_{1c} = p_1 - \frac{\rho L}{2A} \cdot \frac{dq_c}{dt} \quad (4a)$$

and

$$p_{a0} = p_2 + \frac{\rho L}{2A} \cdot \frac{dq_c}{dt} \quad (4b)$$

in which p_{1c} and p_{a0} are the pressures ahead of and behind the valve in the channel. ρ is the density of the liquid, A is the cross-sectional area of the channel and L is the length of the channel. For the system without valve is

$$p_{1c} = p_{a0} = p_k \quad (5)$$

where p_k is the pressure in the channel at the place where, normally, the valve is situated. Similarly to the equations for reservoir 1, those for reservoir 2 are given by

$$\frac{dp_{a2}}{dt} = \frac{1}{C_2} \cdot q_{a2} \quad (6a)$$

$$\text{with} \quad C_2 = Q_{a2} / \gamma \cdot \bar{p}_{a2}. \quad (6b)$$

Furthermore, we suppose

$$q_{a2} = \hat{q}_{a2} \cdot \sin(\omega t + \theta) \quad (7)$$

$$p_2(t) = p_{a2}(t). \quad (8)$$

One important difference in the relation (7), in comparison with (2), is the introduction of the phase angle θ between the pressure variations p_1 and p_2 . The system equations (1) to (8) are solved, using the complex calculation method in which is stated that

$$p_1 = p_0 + \hat{p}_1 \cdot e^{j\omega t} \quad (9a)$$

$$p_2 = p_0 + \hat{p}_2 \cdot e^{j(\omega t + \theta)} \quad (9b)$$

and

$$q_c = \hat{q}_c \cdot e^{j(\omega t + \theta_q)} \quad (9c)$$

Here p_0 is the mean level of p_1 and p_2 and $\hat{p}_1$, $\hat{p}_2$ and $\hat{q}_c$ are the amplitudes of the time varying component of p_1 , p_2 and q_c respectively.

From equations (1)–(3) and (6)–(8) it is found that

$$\hat{p}_1 = -\frac{\hat{q}_{a1}}{\omega C_1} \quad (10a)$$

and

$$\hat{p}_2 = -\frac{\hat{q}_{a2}}{\omega C_2}. \quad (10b)$$

When substituting relation (9) into relations (4) and (5) two complex relations are obtained in which we equalize the real and the imaginary parts respectively. This gives the solution

$$p_k = p_0 + \hat{p}_k \cdot \cos(\omega t + \theta_p) \quad (11a)$$

$$q_c = \hat{q}_c \cdot \cos(\omega t + \theta_q) \quad (11b)$$

with

$$\hat{p}_k = \frac{1}{2} (\hat{p}_1^2 + 2\hat{p}_1 \hat{p}_2 \cos \theta + \hat{p}_2^2)^{1/2} \quad (11c)$$

$$\hat{q}_c = \frac{A}{\omega \rho L} (\hat{p}_1^2 - 2\hat{p}_1 \hat{p}_2 \cos \theta + \hat{p}_2^2)^{1/2} \quad (11d)$$

and

$$\tan \theta_p = \frac{\sin \theta}{a + \cos \theta} \quad (11e)$$

$$\tan \theta_q = \frac{a - \cos \theta}{\sin \theta} \quad (11f)$$

with

$$a = \hat{p}_1 / \hat{p}_2.$$

From this equation it can be seen that the phase angle between flow q_c and pressure p_k ($\theta_p - \theta_q$) follows from the relation

$$(\theta_p - \theta_q) = \arctan \left(\frac{\sin \theta}{a + \cos \theta} \right) - \arctan \left(\frac{a - \cos \theta}{\sin \theta} \right). \quad (12)$$

Theoretical results

Figure 2 shows the system diagram in which the phase difference ($\theta_p - \theta_q$) is calculated as a function of the independent parameters a and θ . In this figure, θ is varied from 0 to -90° in steps of 10° . For the positive values of θ these curves are mirrored along the a -axis. From this figure one can see that for $a = 1$ the flow and the pressure are always in phase, independently of θ . From equation (11d) it can be seen that the value of θ then determines the flow value: at $\theta = 0^\circ$, $q_c = 0$ and at $\theta = 90^\circ$ q_c is maximal.

To create physiological pressures (p_{1c} is between 0 and 16 kPa and p_{a0} is between 8 and 16 kPa), it is necessary to set $a = 2$. Figure 2 shows that for this value of a the minimum phase difference between flow and pressure is about 37° . The phase difference θ between p_1 and p_2 has then to be 90° . Furthermore, it must be noted that ($\theta_p - \theta_q$) is only slightly dependent on θ when θ is in the range of -60 – 90° . For $\theta = 0$ a sudden change in ($\theta_p - \theta_q$) is found for $a = 1$. It may be briefly stated that the best approximation of the physiological condition is found when a is set at 2 and $\theta = -90^\circ$. The phase difference between flow and pressure is then 37° . Figure 3a shows the pressure and flow patterns for this theoretical situation.

The system with a valve

Up to now we have dealt with the system as if it had neither valve nor pump. Now we introduce those components with their corresponding equations. For a correct valve function the following assumptions are made

- (1) the valve opens when $p_1 > p_2$,
- (2) the maximum flow through the valve equals the flow in the system without valve,
- (3) the valve closes when q_c equals an experimentally determined minimum flow q_{vmin} ,
- (4) at the moment of valve closing ($t = t^*$) a supplementary pressure change appears at the aortic side and at the ventricular side of the valve. The magnitude of this peak is

$$\left. \frac{\rho L}{2A} \cdot \frac{dq_c}{dt} \right|_{t=t^*} \quad (13)$$

the contribution of which is positive for the aortic side and negative for the ventricular side.

- (5) After the valve is closed, p_{1c} will follow pressure p_1 and p_{a0} will follow p_2 .

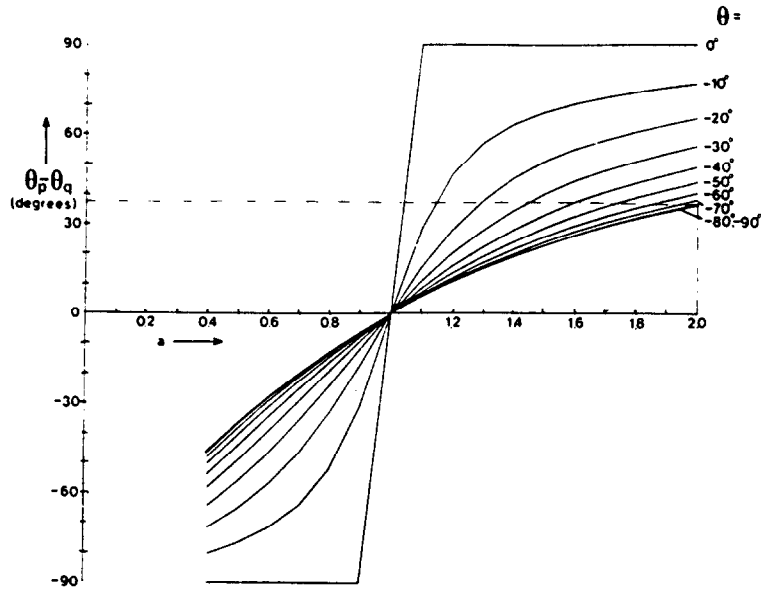


Fig. 2. The theoretically obtained phase difference between pressure p_k and flow q_v as a function of parameters a and θ for the system without valve. θ is varied from 0 to -90° in steps of 10° , while a is varied continuously from 0.4 to 2.0.

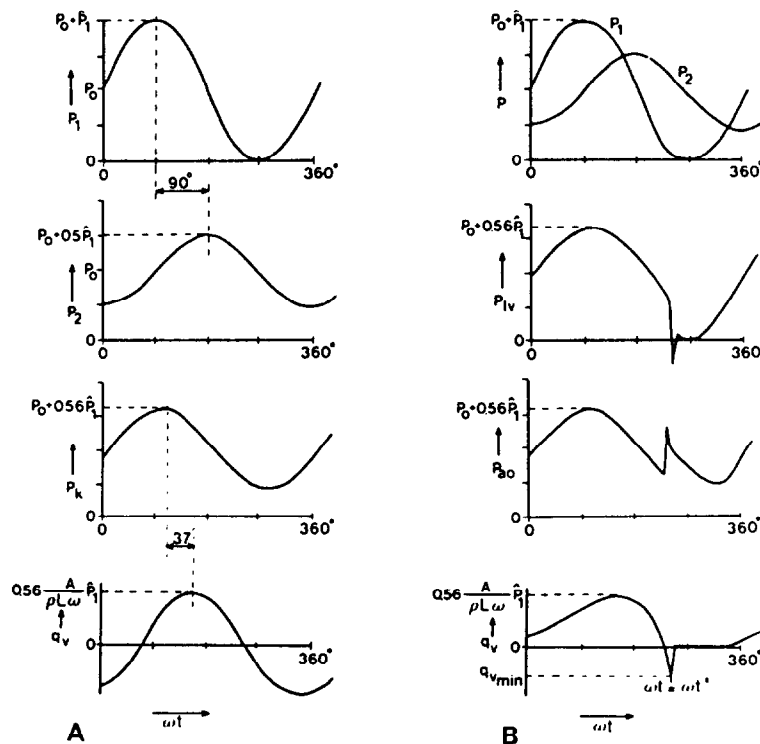


Fig. 3. The theoretically expected patterns of pressure and flow for the situation in which $a = 2$ and $\theta = -90^\circ$. (a) Situation without valve. (b) Situation including valve. Symbols used: p_1 , p_2 : pressure in reservoirs 1 and 2 respectively; p_{1v} , p_{2v} : pressure ahead of and behind the valve; p_k : pressure at the level of valve; q_v : flow through the channel.

Taking these assumptions into consideration, it may be expected that the pressure and flow patterns in the system are well described by the curves in Fig. 3b. Finally, to keep the whole system stationary, the pump has to produce a constant volume flow per second, that is

$$q_p = f \cdot \int_{t=0}^{2\pi} q_t dt \quad (14)$$

with f = frequency of valve loading (Hz).

This value of q_p equals the forward flow through the valve minus the leak flow through the valve. On the other hand, the adjusted value of q_p directly determines the value of the forward flow and therefore of q_v , while it may be expected that it does not affect the leak flow.

Measurements and results

Experiments were performed to verify the theoretical analysis, and more specifically the system diagram of Fig. 2. These experiments were carried out on the system without a valve and with two identical reservoirs (see Fig. 1a). The fluid velocity in the centre of the channel was measured just behind the valve, using an electromagnetic catheter-tip velocity meter which was connected to a sine-wave electromagnetic flow meter with a carrier frequency of 600 Hz and an upper frequency response of 100 Hz. (-3 dB, Transflow 600). The pressures in the liquid of the two reservoirs (p_1 and p_2) were measured through thin needles connected to manometers (Statham P23dB), and the pressures p_{1c} and p_{ao} were measured with Millar catheter-type micromanometers (PC350). These pressures were measured 1.5 cm ahead of and 1.5 cm behind the valve with an upper frequency of 2 kHz. The determined variables were recorded with a multichannel physiological recorder (Schwarzer 80185) and on magnetic tape, using an instrumentation recorder (HP3968A). The two independent parameters which were varied, were the phase difference between p_1 and p_2 (θ) and the ratio of the amplitudes of the time varying component of the pressures p_1 and p_2 ($a = \hat{p}_1/\hat{p}_2$). The system was started after θ was adjusted.

Variation of a was brought about by increasing or decreasing the magnitude of p_1 . This procedure was carried out for five values of θ , namely $\theta = 5, 60, 92, -32$ and -70° . a was varied in the range of 0.4–1.7. This was the widest possible range for this system with two identical reservoirs and without valve. The phase difference between flow and pressure was measured from the signals which were written on paper with the multichannel recorder. The inaccuracy of this measuring procedure is $\pm 5^\circ$. Figure 4 gives the results of these measurements in combination with the theoretically determined system diagram lines appropriate to the adjusted phase differences. From this figure it is concluded that the simple theoretical description of the system is fairly reliable.

Next, experiments were performed in the system with a valve. In order to obtain a fair approximation to the physiological pressure pattern, a must have the value 2 and the mean pressure value in reservoir 2 needs to be raised. This approximation to the physiological condition can be realized by holding the pressure p_1 between 0 and 16 kPa and p_2 between 8 and 16 kPa. To overcome problems in adjusting the system, reservoir 1 is replaced by a smaller one. As a valve, a Björk-Shiley valve (21 ABP) has been taken. Furthermore, a gear-wheel pump (Verder 114–1, ty-316) with a maximum capacity of $6 \text{ l} \cdot \text{min}^{-1}$ was used. From the experiments, it was found that in the system with a valve, the phase difference between p_1 and p_2 has to be adjusted to get a good physiological equivalent. The optimum condition is reached at the following settings: $\theta = -35^\circ$, p_1 ranges between 0 and 16.8 kPa and p_2 ranges between 8 and 16 kPa. The results of the experiment with this optimum condition are given in Fig. 5a. From this figure it may be concluded that the pressure and flow patterns qualitatively mimic the shapes of the *in-vivo* curves, while the pressure and flow curves are closely in phase. Increasing the phase difference (up to $\theta = -45^\circ$) results in a valve closure which is too early (see Fig. 5b), while decreasing the phase difference (up to $\theta = -22^\circ$) results in a too late valve closure (Fig. 5c). Further, from Fig. 5 it is seen that in the patterns of p_1 and p_2 an oscillation is present when the valve closes. This is probably caused by the needles used in measuring the pressures p_1 and p_2 . In the signals of p_{1c} and p_{ao} ,

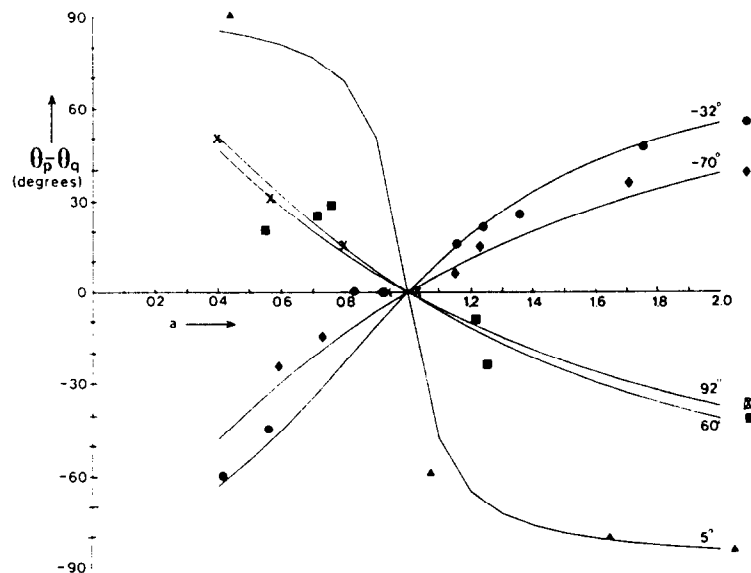


Fig. 4. Experimental verification of the model for the system without a valve. For five values of θ and a ranging between 0.4 and 1.8, the theoretically predicted phase angles between pressure p_k and flow q_t in the middle of the channel ($\theta_p - \theta_q$) are verified experimentally. The measured points are denoted by the following symbols: $\theta = -32^\circ$: $\bullet$; $\theta = -70^\circ$: $\blacklozenge$; $\theta = 92^\circ$: $\times$; $\theta = 60^\circ$: $\blacksquare$; $\theta = 5^\circ$: $\blacktriangle$. The measurement accuracy of $\theta_p - \theta_q$ is $\pm 5^\circ$.

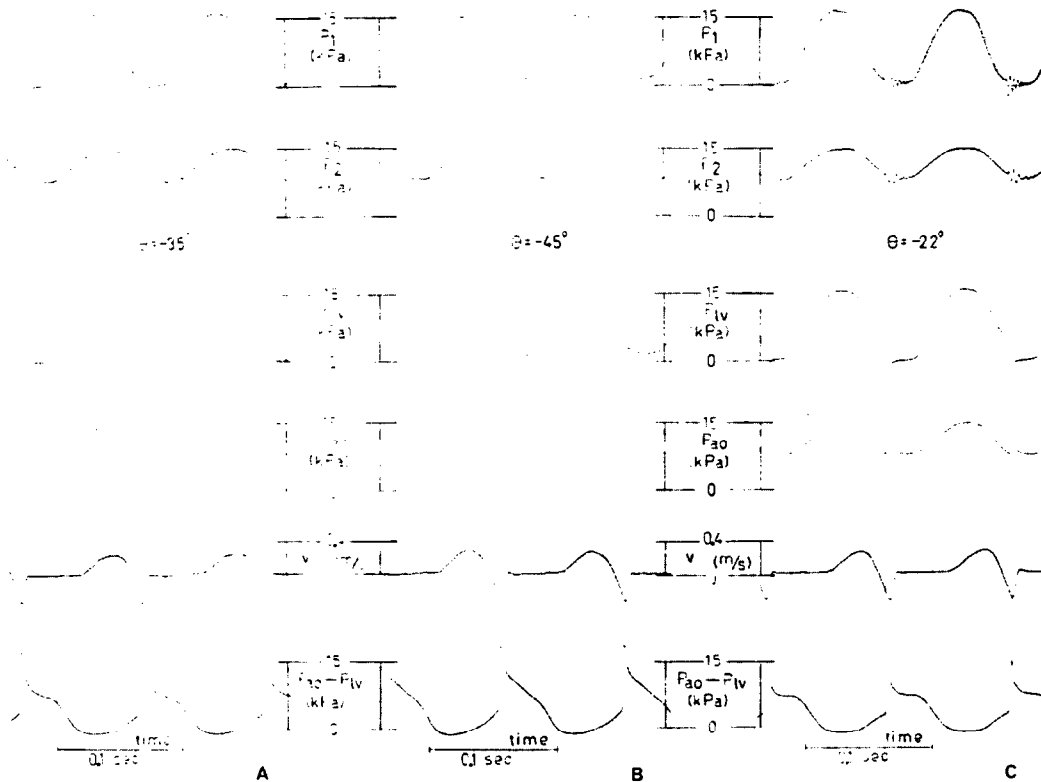


Fig. 5. Measured signals in the system including a Björk-Shiley valve prosthesis (21ABP) with a closing-and-opening frequency of 10 Hz. (a) The curves for the system in optimum condition (phase difference between pressure and flow is minimal) ($\theta = -35^\circ$). (b) Influence of increasing θ ($\theta = -45^\circ$). The valve now closes too early. (c) Influence of decreasing θ ($\theta = -22^\circ$). The valve now closes too late.

measured with the Millar catheters, the oscillations are absent. Just after the valve is closed, a pressure peak appears at the aortic side and at the ventricular side of the valve. Theoretically the magnitude of these peaks is given by relation (13). For the optimum case ($\theta = -35^\circ$), q_v was determined from the measured fluid velocity at valve closure multiplied by the cross-sectional area, which lead to a magnitude of 3.6 kPa. From Fig. 3b it can be seen that this peak at the aortic side amounts to about 4 kPa and at the ventricular side amounts to about 5 kPa. Finally, the results for a Hancock bioprosthesis (242-A21) are given in Fig. 6. Figure 6a shows the tracings: owing to the compliance of the valve leaflets the pressure peaks have been diminished. A frontal view of the valve was obtained with an endoscope inserted through reservoir 2. In Fig. 6b photographs of this valve during an accelerated test are shown, illustrating that the valve closes completely.

DISCUSSION

In this paper a system is described in which heart valve prostheses can be physiologically loaded at a frequency of 10 Hz, although the system can also be used at higher or lower frequencies. In the described system this requirement can be met by the many control possibilities. It is possible to change the pressure patterns p_1 and p_2 in magnitude and in phase difference. The flow through the system can be adjusted by a pump. The heart valve prosthesis, which is loaded, can be observed and photographed or filmed during the whole test.

The number of cycles gone through by the valve is registered. Furthermore, a built-in safety system switches off the whole system when something goes wrong, for example when the valve prosthesis is damaged or a tube gets loose. No long term problems with the heat production have to be expected as the discs in the control mechanisms rotate quite slowly. This is furthermore fortified by the fact that the rotating discs cool automatically by the air constantly flowing in and out through the control mechanisms. Moreover, only a small amount of liquid is necessary. The system has been analysed theoretically as well as for the case with and without a valve inserted.

The results of the measurements and the theoretical analysis for the system without a valve show a fair agreement. For the system with a valve, the mean pressure in the downstream reservoir has to be raised in comparison with the mean pressure value in the upstream reservoir to provide physiological circumstances, which is not accounted for in the model. But also here the experimentally observed flow pattern qualitatively resembles the predicted one. This suggests that the simple physical model, based on a lumped circuit approach and the neglect of the effects of heat conduction in the reservoir, fluid viscosity in the channel and local flow phenomena at the connection of reservoir and channel, holds for the adjusted high frequencies. Furthermore, due to the high frequency and the short inlet length, the assumption is justified that the velocity profile in the channel is flat. However, these assumptions can not be made at low frequencies which restricts the applicability range of the system.

In conclusion it is stated that the developed apparatus may

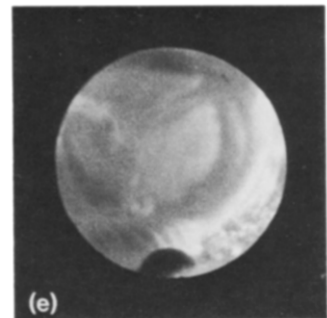
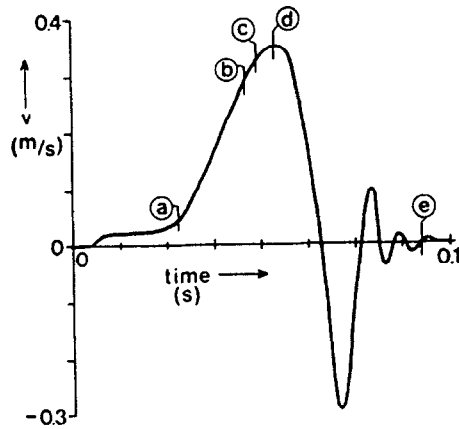
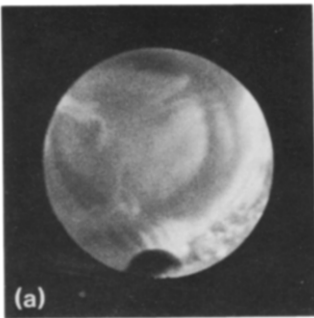
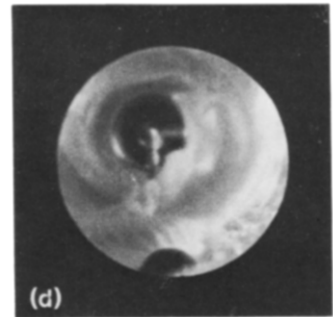
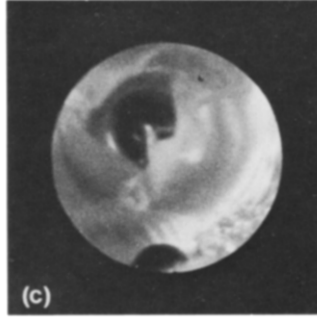
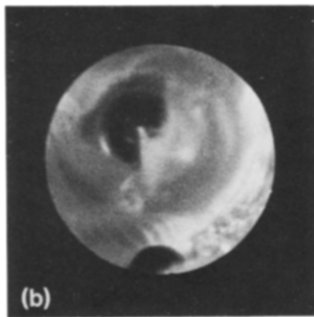
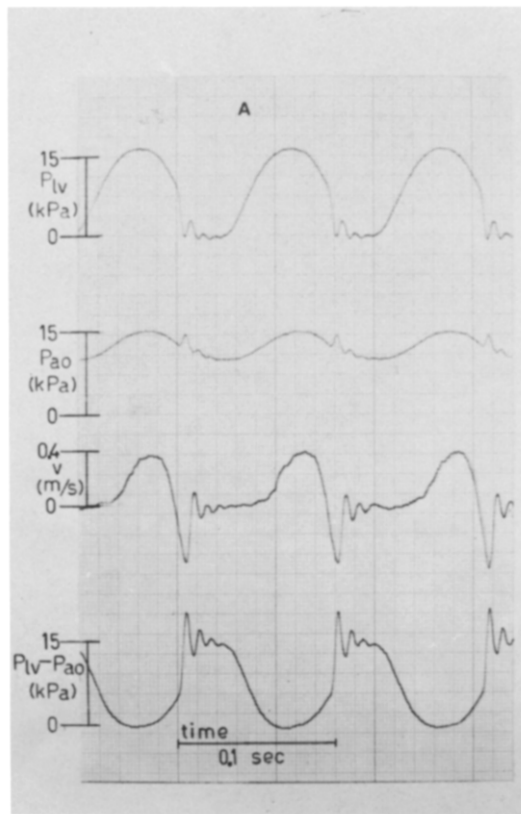


Fig. 6. Measurements made in a system including a Hancock bioprosthesis (242, Aortic, 21). (A) Tracings measured with a Hancock valve. ($\theta = -35^\circ$). (B) Photographs of the valve during an accelerated test. For the cases (a) and (e) the valve is closed and for the cases (b), (c) and (d) the valve is opened.

be expected to be a useful tool in the accelerated evaluation study of heart valve prostheses.

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REFERENCES

- Almond, C. H., Mayhan, K. G., Young, R. D., Simmons, E. M., Patterson, B. R., Biolsi, M. E., Davis, S. A. and McClatchy, B. J. (1974) A physiological approach to high-frequency testing of prosthetic ball valves. *J. thorac. cardiovasc. Surg.* **67**, 839–848.
- Clark, R. E., Swanson, W. H., Kardos, J. L., Hagen, R. W. and Beauchamp, R. A. (1978) Durability of prosthetic heart valves. *Ann. thorac. Surg.* **26**, 323–335.
- Häußinger, G. and Reul, H. (1981) Konstruktion und Aufbau eines Dauerfestigkeitsprüfstandes für künstliche Herzklappen. *Biomed. Technol.* **26**, 13–15.
- Haworth, W. S., Lewis, R. W. H., Bellhouse, B. J. and Bellhouse, F. H. (1976) Design and use of a new machine for accelerated fatigue testing of prosthetic heart valves. *Proc. Eur. Soc. artif. Organs* **3**, 138–142.
- Koorajian, S., Keller, D. P., Pierce, W. R., Starr, A. and Kerr, R. (1969) Criteria and systems for testing artificial heart valves *in-vitro*. *Prosthetic Heart Valves, Proceedings of the 2nd National Conference*, (Edited by Brewer), pp. 92–112. Thomas, Springfield.
- Martin, T. R. P., van Noort, R., Black, M. M. and Morgon, J. (1980) Accelerated fatigue testing of biological tissue heart valves. *Proc. Eur. Soc. artif. Organs* **7**, 315–319.
- Quinton, W. E., Afstun, M. S., Leyse, R. M., Wintersheid, L. C., Harrison, A. G. and Merendino, K. A. (1961) An accelerated fatigue pump for testing prosthetic aortic valves from *Prosthetic Valves for Cardiac Surgery*, (Edited by Merendino, K. A.), pp. 234–243. Thomas, Springfield.
- Rousseau, E. P. M., Sauren, A. A. H. J., van Hout, M. C. and van Steenhoven, A. A. (1983) Elastic and viscoelastic material behaviour of fresh and glutaraldehyde treated porcine aortic valve tissue. *J. Biomechanics* **16**, 339–349.
- Sauren, A. A. H. J., Kuypers, W., van Steenhoven, A. A. and Veldpaus, F. E. (1980) Aortic valve histology and its relation with mechanics, preliminary report. *J. Biomechanics* **13**, 97–104.
- Sauren, A. A. H. J., van Hout, M. C., van Steenhoven, A. A., Veldpaus, F. E. and Janssen, J. D. (1983) The mechanical properties of porcine aortic valve tissue. *J. Biomechanics*, **16**, 327–338.
- van Steenhoven, A. A. and van Dongen, M. E. H. (1979) Model studies of the closing behaviour of the aortic valve. *J. Fluid Mech.* **90**, 21–32.
- van Steenhoven, A. A., Verlaan, C. W. J., Veenstra, P. C. and Reneman, R. S. (1981) An *in-vitro* cinematographic analysis of the behaviour of the aortic valve. *Am. J. Physiol.* **240**, 286–292.
- van Steenhoven, A. A., van Duppen, Th. J. A. G., Cauwenberg, J. W. G. and van Renterghem, R. J. (1982a) *In-vitro* closing behaviour of Björk-Shiley, St Jude and Hancock heart valve prostheses in relation to the *in-vivo* recorded aortic valve closure. *J. Biomechanics* **15**, 841–848.
- van Steenhoven, A. A., Veenstra, P. C. and Reneman, R. S. (1982b) The effect of some hemodynamic factors on the behaviour of the aortic valve. *J. Biomechanics* **15**, 941–950.
- Steinmetz, G. P., May, K. J., Mueller, V., Anderson, H. N. and Merendino, K. A. (1964) An improved accelerated fatigue machine and pulse simulator for testing and developing prosthetic cardiac valves. *J. thorac. cardiovasc. Surg.*, **47**, 186–198.
- Swanson, W. M. and Clark, R. E. (1978) Flow and pressure measurements in accelerated fatigue testing of prosthetic heart valves. *Ann. thorac. Surg.*, **26**, 323–335.