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DEVELOPMENT OF A HYDRAULIC ANALOG OF THE HUMAN CIRCULATORY SYSTEM FOR TESTING ARTIFICIAL HEARTS

1. Parameter Optimization of the Hydraulic Model Elements

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ABSTRACT

For the application of artificial hearts or artificial heart valves, a hydraulic analog model of the circulatory system must be developed to check and evaluate their performances prior to animal and clinical experiments. In this study, parameter estimation of the hydraulic model components has been conducted by means of an electrical analog simulation. This analog model can be represented by linear four-terminal networks consisted of the according number of resistances, inductances and capacitances for the systemic and pulmonary circulations. Parameter optimization has been governed in respect to the correct pressure-flow relations, correct input impedances and correct flow work rates of the model. Results of the estimation lead to the suitable numerical values for the actual design of the hydraulic model.

1. Introduction

For the evaluation of performance and function of artificial hearts or artificial heart valves, it must be tested in a hydraulic analog model of the circulatory system prior to their implantation which consumes a time and expensive animal preparations. In such a mock-circulation, several parameters of the circulation can be kept constant over a desired period of time or be varied at choice, and it is very useful to determine the cardiac function curves (cardiac output v.s. atrial pressure) and the involved control mechanisms. But it must be postulated generally and qualitatively that the model simulates physiological conditions in a sufficient approximation. It means that the model is geometrically similar, the instationary flow processes are similar to those of physiological data and the model fluid has a suitable viscosity.

In testing artificial hearts, it is of prime concern to get informations about the ejection and filling behavior and the correlated flow work rate, which both are governed by the instationary flow process at the four in-and outlets of the artificial heart. These instationary flow phenomena are strictly correlated to the input and output impedance of the circulatory system, defined as the ratio of oscillatory pressure to oscillatory flow. Thus, it follows subsequently that, for the described purpose, the model is to be developed, which approximates the impedances of both the systemic and pulmonary circulations pertinent for an artificial heart.

Westerhof et al. (1971) developed a hydraulic arterial system for pumping hearts and showed that a simple model consisting of two laminar resistances and an adjustable compliance represents the input impedance of a real arterial system in good approximation. We have modified and extended this model, striving for a simulation of the venous and atrial components as well.

This paper is concerned with a hydraulic equivalent model of the circulatory system for testing artificial hearts or artificial heart valves. It will be shown that it is possible to arrive at an easily adjustable hydraulic load with the input impedances and flow work rates equal to those of both systemic and pulmonary circulations. Parameter estimation for the hydraulic load components will be conducted by means of an electrical analog model. Basing on the parameter estimation, the design and the construction of a hydraulic model can be done. The analog model will be discussed concerning with the input impedances and the flow work rates, and compared with those of physiological data.

2. Model Concept of the Circulatory System

We have approximated the human systemic and pulmonary circulatory systems each by a symmetrical 9-component hydraulic model, consisted of laminar resistances and compliances with the inherent inertance parts (fig. 1). Each component, $Z_2 \sim Z_7$ represents a lumped property of a certain region of the vascular system. This rough correlation is listed in Table 1. All of the listed elements have additional inertance components which are unaviodable in the design of a hydraulic model, but these introduced inertances have no physiological correlation at all. The components Z_0 and Z_8 represent resistance and inertia of rigid flow-probes at the inlet and outlet of the system, which must be incorporated in the analysis. Especially the flow-probe at the inlet has a significant influence on the impedance characteristics of

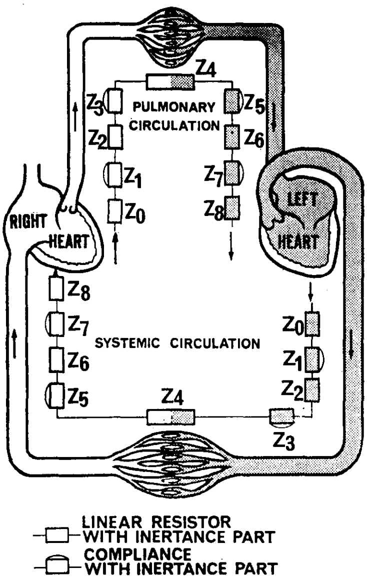

Fig. 1. Human circulatory system and its comparison with the lumped hydraulic model

Table 1. Anatomical or functional equivalent of the model component

Number of Element	Type of Element	Anatomical or Functional Equivalent
Z_2	Linear Resistor	Characteristic resistance of systemic respectively pulmonary arteries
Z_3	Compliance	Compliance of systemic respectively pulmonary arteries
Z_4	Linear Resistor	Peripheral resistance of systemic respectively pulmonary circulation
Z_5	Compliance	Compliance of the venous system of systemic respectively pulmonary circulation
Z_6	Linear Resistor	Characteristic resistance of the venous system of systemic respectively pulmonary circulation
Z_7	Compliance	Compliance of the right respectively left atrium

the model. Because of its inherent high hydraulic inertance, modulus and phase of the input impedance will be shifted to high values and to positive phase angles especially in the range of the higher harmonics, respectively. This effect, however, can be compensated by an additional compliance element Z_1 coupled to the flow-probe.

For the model construction it will be useful to elucidate some common features of vascular input impedance in mammals. There exists a broad range of possible vascular impedance in animals, but some general features of the impedance spectra are similar in any case. The absolute value of the input impedance (Z) is high for steady flow and is equal to the peripheral resistance, the phase is, by definition zero. For low frequencies in the range of 2 or 3 Hz, Z decreases to a low value and then remains nearly constant for higher frequencies. The low Z value is generally agreed to be the characteristic impedance of the proximal vessel where the input impedance is measured. The phase of the input impedance is negative at low frequencies and about zero or positive for higher frequencies. This qualitative description of the input impedance is correct for all major beds of the vascular systems (PATEL et al. 1963, GABE et al. 1964, TAYLOR 1966, O'ROURKE 1968, WESTERHOF et al. 1971, GESNNER 1972).

2.1 Electrical Analog of the Hydraulic Model

Basing on these considerations, first, an electrical analog for the estimation of numerical design of the hydraulic elements has been developed. This model can be described by an electrical four terminal network with the according number of resistances, compliances and inductances. Fig. 2 shows the hydraulic and the equivalent electrical analog model, from which the following equations are obtained:

$$P_0 - P_1 = L_0 \frac{df_0}{dt} + R_0 f_0 \tag{1}$$

$$P_1 = \frac{1}{C_1}(V_1 - V_{u1}) \tag{2}$$

$$V_1 = V_{10} + \int (f_0 - f_2)dt \tag{3}$$

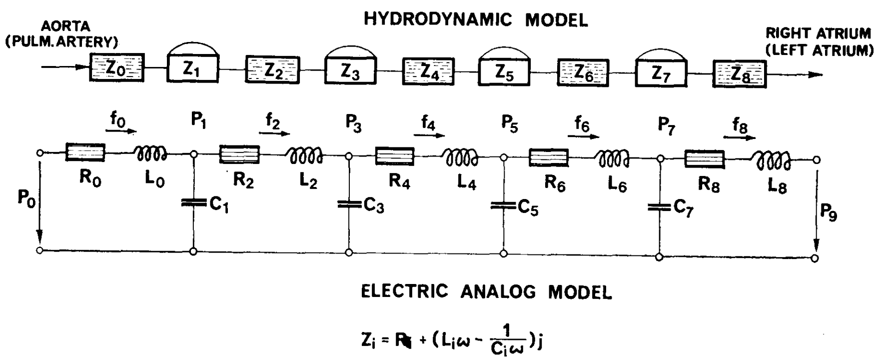

Fig. 2. Comparison of the hydraulic and the electrical analog model of human circulatory system

$$P_1-P_3=L_2\frac{df_2}{dt}+R_2f_2 \tag{4}$$

$$P_3=\frac{1}{C_3}(V_3-V_{u3}) \tag{5}$$

$$V_3=V_{30}+\int(f_2-f_4)dt \tag{6}$$

$$P_3-P_5=L_4\frac{df_4}{dt}+R_4f_4 \tag{7}$$

$$P_5=\frac{1}{C_5}(V_5-V_{u5}) \tag{8}$$

$$P_5-P_7=L_6\frac{df_6}{dt}+R_6f_6 \tag{10}$$

$$P_7=\frac{1}{C_7}(V_7-V_{u7}) \tag{11}$$

$$V_7=V_{70}+\int(f_6-f_8)dt \tag{12}$$

$$P_7-P_9=L_8\frac{df_8}{dt}+R_8f_8 \tag{13}$$

V_{ui} and V_{i0} represent the zero pressure volume (the unstressed volume) and the initial volume, which both can be set at reasonable values of interest. These values affect only on the volume terms since the compliance elements are linear. Parameter estimation for the hydraulic elements of systemic and pulmonary circulation has been carried out by means of an analog computer (EAI Model 180). The inlet and outlet pressure (voltage) curves (P_0 and P_9) are generated by programmable function generators. Shape and absolute values of these curves are generated according to physiological curves from a literature (GREGG 1961). The impedance spectra of the analog model has been measured by utilizing a sinusoidal function generator (Hewlett Packard Model 3310A) and a frequency response analyser (Schlumberger Model 1312).

2.2 Parameter Estimation of the Hydraulic Elements

The flow (current) curves and the impedance spectra are optimized in an iterative process by suitable variation of the network components and by continuous comparison of the resulting data with those from in-vivo measurements. This parameter estimation process has been governed by the following criteria:

1) correct absolute values of flow maxima and minima,
2) correct systolic and diastolic duration of the flow curves,
3) correct shape and zero-level of the flow curves,
4) correct impedance moduli,
5) correct impedance phases,

6) the resulting parameter values should as far as possible be physiologically meaningful and be in the right order of magnitude, and

7) the parameter values of hydraulic analog elements should be realizable for the construction.

The application of these intermeshed criteria guarantees that the obtained parameter values are unique within very narrow limits. The obtained set of optimum parameter values for the hydraulic model of systemic and pulmonary circulation is listed in Table 2.

Table 2. Set of optimum parameter values for the simulation of systemic and pulmonary circulations

Element	Systemic Circulation	Pulmonary circulation	Dimension
R_0	0	0	dyn sec/cm^5
L_0	1.5	1.5	dyn sec^2/cm^5
C_1	$0.165 \cdot 10^{-3}$	$0.067 \cdot 10^{-3}$	cm^5/dyn
R_2	90	45	dyn sec/cm^5
L_2	1.1	1.0	dyn sec^2/cm^5
C_3	$1.1 \cdot 10^{-3}$	$2 \cdot 10^{-3}$	cm^5/dyn
R_4	1300	110	dyn sec/cm^5
L_4	3.1	2.4	dyn sec^2/cm^5
C_5	$1 \cdot 10^{-2}$	$3 \cdot 10^{-2}$	cm^5/dyn
R_6	25	25	dyn sec/cm^5
L_6	1.7	1.7	dyn sec^2/cm^5
C_7	$5 \cdot 10^{-3}$	$5 \cdot 10^{-3}$	cm^5/dyn
R_8	0	0	dyn sec/cm^5
L_8	1.1	1.1	dyn sec^2/cm^5

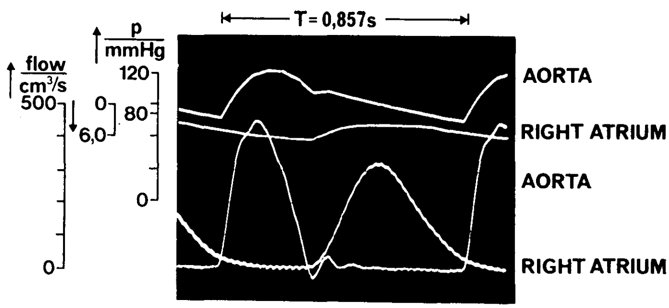

Fig. 3. Pressure- and flow curves at the in- and outlet of the systemic circulation, Aorta: P_0, f_0, Right atrium: P_9, f_8

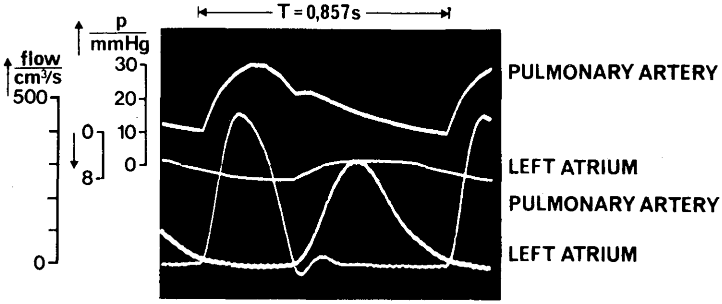

Fig. 4. Pressure- and flow curves at the in- and outlet of the pulmonary circulation, Pulmonary : P_0, f_0, Left atrium : P_9, f_8

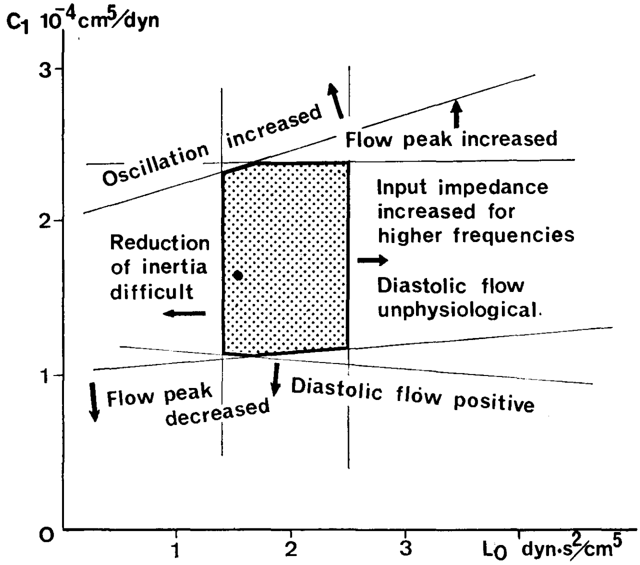

Fig. 5. Optimum range for the compliance and inertance of inlet part of the systemic circulation model

The corresponding set of pressure and flow curves is represented in the next two figures (Fig. 3 and Fig. 4). The pressure curves are preset and the flow curves result at the above tabulated model parameters. The represented figures demonstrate that the tidal course as well as the absolute values of the obtained flow curves are sufficiently acceptable. It should be particularly noted here, that the active contraction of atria is not taken into account. (And note that the atrial pressure curves are inverted only because of technical reasons.) But this is a realistic omission, since an artificial heart has no active atria. Accordingly the atrial pressure

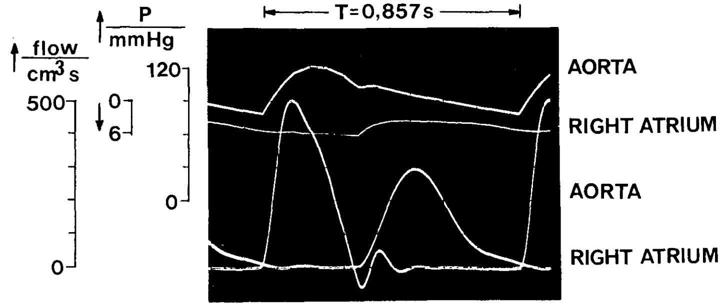

Fig. 6. Pressure- and flow curves at the in- and outlet of the systemic circulation for large inlet compliance: $C_1=3\times15^{-4}cm^5/dyn$

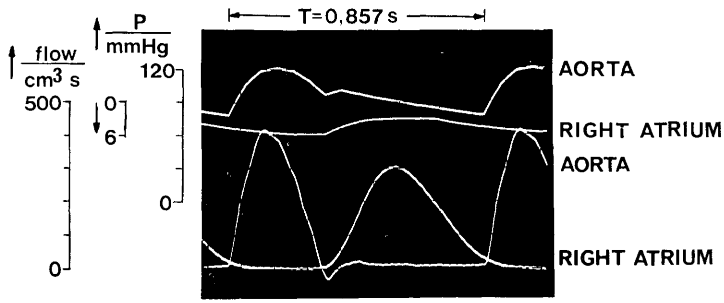

Fig. 7. Pressure- and flow curves at the in- and outlet of the systemic circulation for small inlet compliance: $C_1=0.825\times10^{-4}cm/dyn$

curve P_9 is represented by a smoothing course of voltage, so that the outlined outflow f_8 may mimic the realistic inflow condition for an artificial heart.

The hydraulic property of each model element dominates characteristics of the pressure-flow curves and impedance spectra of the system. In particular, properties of the inlet part (L_0, C_1) significantly affect pressure-flow process and input impedance. Fig. 5 shows schematically different criteria of the optimization process, which represents optimum range of L_0 and C_1 for the systemic circulation. At the same time it is to be seen that only a very limited range of the inertance and compliance represented here by the shaded area, is suitable for design of the hydraulic analog model. All sets of parameter outside of this range lead to the indicated difficulties. The design point of our model is depicted within the shaded area. Fig. 6 and Fig. 7 show the pressure-flow curves for $C_1=3\times10^{-4}$ cm/dyn and for $C_1=0.825\times10^{-4}$ cm^5/dyn, respectively in the systemic circulation. Increasing of the compliance results in that the peak value of aortic flow and the backward-flow at the end of systole increases. In the case that the compliance decreases, the peak value of aortic flow decreases, and the positive diastolic flow can be seen. As shown in Fig. 6, the aortic flow curve is superimposed with a slight oscillation during diastole. This phenomenon also can be observed when the inertance of inlet part L_0 decreases, likeweise when the inertance L_2 increases. Further considerations will be discussed later.

2.3. Impedance Spectra

Vascular input impedance can be defined as the ratio of oscillatory inlet pressure to the consequent oscillatory flow, and calculated by means of Fourier analysis method. This definition holds only for sinusoidal waveforms, both in the same frequency. Though the following considerations are not pertinet for our analog set-up which is linear and periodic, it should be pointed out that the application of Fourier analysis is restricted only to linear systems and to periodic phenomena. Both prerequisites are not fulfilled in the real circulatory system, since neither the cardiac action is strictly periodic nor the cardiovascular system is linear. ATTINGER et al. (1966) however pointed out that the mentioned two basic postulates for the Fourier analysis are approximately satisfied within the circulatory system and errors introduced by deviations from these two conditions are in the range of measurement errors.

Hence, the pressure and flow curves can be decomposed into a series of sinusoidal components and represented by a sum of the following forms:

$$\begin{aligned} P(t) &= P_0 + \Sigma\,(A_k \sin kwt + B_k \cos kwt) \\ &= P_0 + \Sigma\, P_k \sin(kwt + \varphi_k) \end{aligned} \tag{14}$$

$$\begin{aligned} f(t) &= f_0 + \Sigma\,(C_k \sin kwt + D_k \cos kwt) \\ &= f_0 + \Sigma\, f_k \sin(kwt + \psi_k) \end{aligned} \tag{15}$$

where P_0 = mean pressure component

f_0 = steady flow component

P_k, φ_k = pressure modulus and phase of k-th harmonic

f_k, ψ_k = flow modulus and phase of k-th harmonic

w = fundamental frequency, i.e. $w=2\pi/T$

T = period of one cardiac cycle,

and

$$P_k = \sqrt{(A_k)^2 + (B_k)^2} \tag{16}$$

$$\varphi_k = \tan^{-1}(B_k/A_k) \tag{17}$$

$$f_k = \sqrt{(C_k)^2 + (D_k)^2} \tag{18}$$

$$\psi_k = \tan^{-1}(D_k/C_k) \tag{19}$$

From the above definitions the impedance Z_k can be represented as follows,

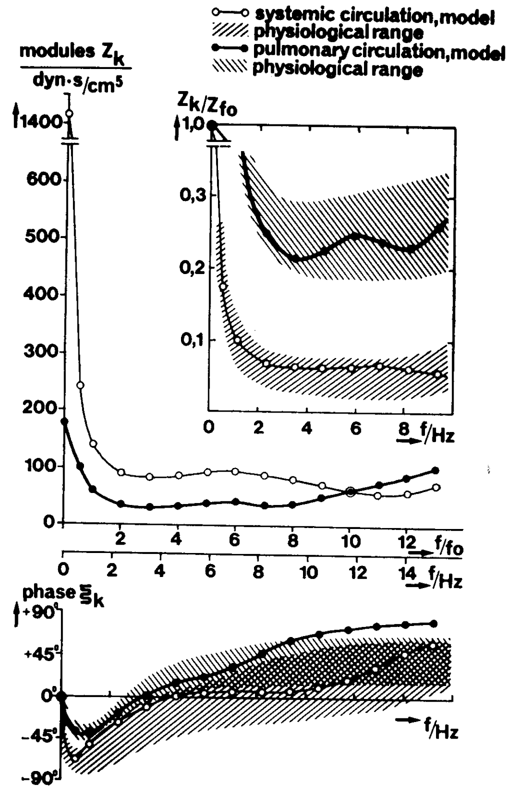

Fig. 8. Input impedance spectra of both the systemic and pulmonary circulations (electrical analog model) compared with physiological data

$$\text{Modulus} \quad Z_k = P_k/f_k \tag{20}$$

$$\text{Phase} \quad \xi_k = (\varphi_k - \psi_k) \tag{21}$$

The impedance for mean pressure and steady flow (Z_0) is the ratio of P_0 to f_0:

$$Z_0 = P_0/f_0 \,. \tag{22}$$

The input impedance spectra for both systemic and pulmonary circulation, as obtained from the described analog model, are depicted in Fig. 8. The hatched areas indicate the physiological bandwidth, which are calculated by the above defined method. The achieved model data are well in this range except phase for the pulmonary circulation above 10 Hz. As mentioned, the impedance moduli are high at very low frequencies and about equals to the peripheral resistance for steady flow. It drops steeply over the first few harmonics and remains nearly constant at the higher harmonics with relatively small fluctuations. It is generally agreed that

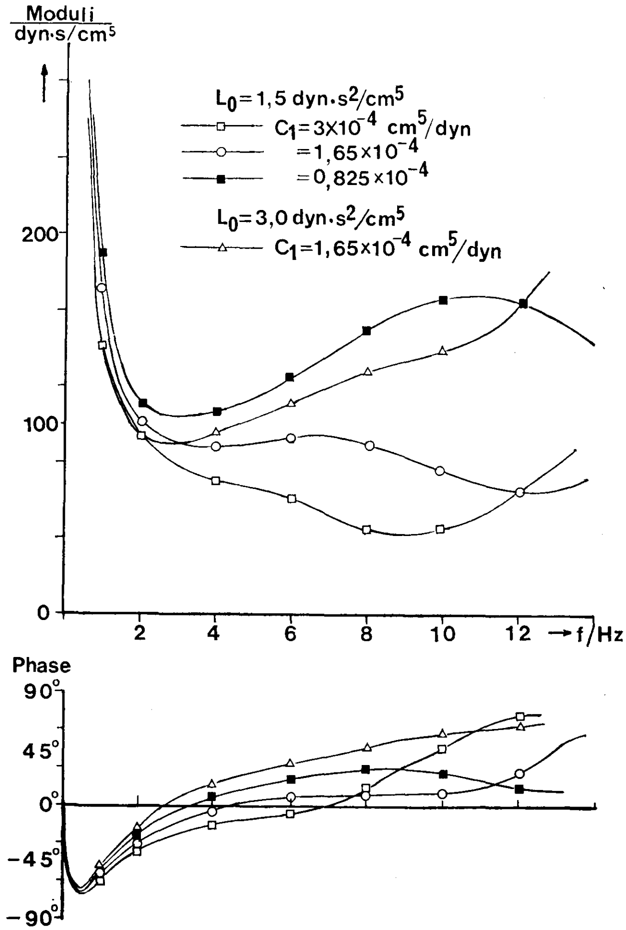

Fig. 9. Effect of the inlet compliance C_1 and inertance L_0 on the input impedance spectra for systemic circulation

this low value of Z is about equal to the characteristic resistance (WETTERER and KENNER 1968) of the coupled arterial bed. The first minimum of Z is found at 2–4 Hz in the aorta. The phase is, by definition, zero at dc, negative at low frequencies (2–3 Hz) and close to zero or positive at higher frequencies. The input impedance of the pulmonary circulation shows the same characteristics as that of the aorta (MILNOR et al. 1969, MILNOR 1972). The model impedance are in acceptable conformity with the characteristics of in-vivo measured impedances.

In the case of the hydraulic properties of the model elements changed, the input impedance of the system also changes, and under certain conditions unphysiological impedance spectra may be observed. Particularly, the properties of inlet part of the system result in various input impedances. Fig. 9 shows the characteristics of the input impedances for the systemic circulation in which the inertance L_0 and compliance C_1 are considered as parameters. The imepedance spectra for $C_1=3\times10^{-4}$cm^5/dyn also describes the acceptable conformity with physiological data. But, as before-mentioned, undesirable superimposed oscillations on the pressure-flow curves can be seen so that it should not use the high compliance in the design of hydraulic analog model. Although the low compliance $C_1=0.825\times10^{-4}$ cm^5/dyn results in the better phase characteristic over the whole of frequency, input impedances are increased in the intermediate frequency range. Increasing of the inertance L_0 to 3.0 dyn sec^2/cm^5 results in the unphysiogical diastolic flow and the input impedance increasing at higher frequencies, which does not agree with physiological data. Reduction of the inertance L_0 is limited by construction factors inherent to the design and results in an extraordinary oscillatory behavior on the pressure-flow curves.

2.4 Input Flow Work Rate

An external work is necessary for a heart to maintain a required blood flow at a certain pressure in the vascular system. Impedance of the vascular system determines afterload of the artificial heart, since the external work of heart can be represented by the product of pressure and flow. The external work of heart per unit time for one cardiac cycle is defined as the flow work rate or the hydraulic power delivered to the system by the following forms:

$$W_T=\frac{1}{T}\int_0^T P(t)f(t)dt \tag{23}$$

which may be rewritten as

$$\begin{aligned}W_T&=\frac{1}{T}\int_0^T[P_0+\Sigma\, P_k\, \sin(kwt+\varphi_k)][f_0+\Sigma\, f_k\, \sin(kwt+\phi_k)]dt\\&=\frac{1}{T}\int_0^T[P_0f_0+\Sigma\Sigma\, P_jf_k\sin(jwt+\varphi_j)\, \sin(kwt+\phi_k)\\&\quad+P_0\,\Sigma\, f_k\sin(kwt+\phi_k)+f_0\,\Sigma\, P_j\sin(jwt+\varphi_j)]dt\\&=\frac{1}{T}\int_0^T\Big[P_0f_0+\frac{1}{2}\,\Sigma\Sigma\, P_jf_k\{\cos[(j-k)wt+\varphi_j-\phi_k]\\&\quad-\cos[(j+k)wt+\varphi_j+\phi_k]\}+P_0\,\Sigma\, f_k\sin(kwt+\phi_k)\\&\quad+f_0\,\Sigma\, P_j\sin(jwt+\varphi_j)\Big]dt\end{aligned} \tag{24}$$

Integration of the 3rd and 4th terms for one cycle gives zero, and integration of the 2nd term gives also zero only if j is not equal to k. Hence, the total flow work rate delivered in unit time comprises the steady term (W_0) and the terms for all harmonics (W_k):

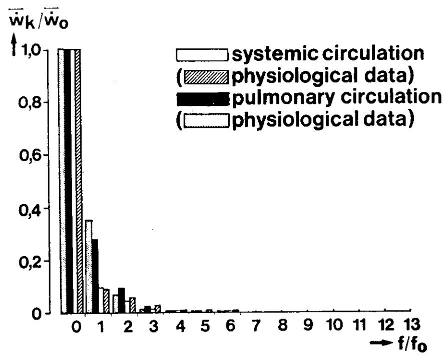

Fig. 10. Frequency spectra of the input flow work rate for both the systemic and pulmonary circulations compared with physiological data

$$\begin{aligned} W_T &= \frac{1}{T}\int_0^T P_0 f_0 + \frac{1}{2}\sum P_k f_k \cos(\varphi_k - \psi_k) \\ &= P_0 f_0 + \frac{1}{2}\sum P_k f_k \cos(\varphi_k - \psi_k) \\ &= P_0 f_0 + \frac{1}{2}\sum P_k f_k \cos \xi_k \\ &= W_0 + \sum W_k \end{aligned} \tag{25}$$

Maintaining the blood flow into vessels, the kinetic power is also needed to deliver into the system. It means that the power input must be represented by the sum of flow work rate and kinetic energy of the blood flowing into the vessels. The above calculation neglects the kinetic power term which, however, is in the range of only 6% of the total power input for the pulmonary circulation (MILNOR et al. 1966) and even smaller for the systemic circulation.

By using the above equation (25), the frequency spectra of the input flow work rates have been calculated for both the model and in-vivo measured data (Fig. 10). The moduli of the single harmonics are normalized with respect to the steady term (W_0) to allow for a better comparison between the spectra. The spectra are again in an acceptable conformity, thus it demonstrates applicability of the model. In both the model and the real systemic arterial system, the pulsatile components (W_k) represent about 15% of the total flow work rate. For the pulmonary circulation, the pulsatile terms are more important and reach a magnitude of about 35% of the total input.

3. Discussion

As shown above, parameter estimation for the hydraulic analog model of the systemic and pulmonary circulations has been conducted by means of an electrical analog model. For the estimation of numerical design of hydraulic elements, linear electrical network models have been used. But there exist actually nonlinear hydraulic properties in a real circulatory system of mammals and in the hydraulic analog models as well. Hitherto, a large number of electrical analog models of human circulatory system have been reported, most of them were investigated under linear electrical network simulation. For the numerical treatment of such a system, an electrical analog is very convenient, because electrical network analysis can be adapted easily, so that it brings available informations for the system, even using the linear simulation. The experimental results illustrated in Fig. 3-4 (pressure-flow curves) and also in Fig. 8 and 10 (input impedances and flow work rates) show sufficient evidences, allowing to use the linear analog model.

They are in acceptable conformity with characteristics of the in-vivo measured data.

The hydraulic property of the model element dominates characteristics of the pressure-flow patterns and the input impedances. Effects of the compliance and inertance at the inlet part of the system have been already explained in Chap. 2.2 and 2.3. Further considerations of those effects for another hydraulic elements will be discussed here as follows:

1. when the characteristic resistance of artery R_2 increases, the peak value of aortic flow (f_0) decreases and pressure P_1 also slightly decreases. When R_2 decreases, the peak value of f_0, the backward-flow of f_0 at the end of systole and pressure P_1 increase. The amplitude and phase of the input impedance show very different aspects over intermediate frequency range (2−12 Hz) even when R_2 changes a small amount (70−110 dyn sec/cm^5). Such a result is also obtained on the occasion of the inertance L_2 changing.

2. Increasing of the compliance C_3 leads to the peak value increasing of the aortic flow, negative diastolic aortic flow and the decreasing of pressure P_3. On the contrary, decreasing of the compliance C_3 reduces the peak value of f_0 and the backward-flow of f_0 at the end of systole, and leads to the positive diastolic flow. The amplitude and phase of the input impedance are not so much dominated by the compliance value and the peripheral resistance R_4 as well.

3. Peripheral resistance R_4, however, gives a change of the steady term of the input impedance (d-c modulus), thus it leads to the basic level shift of aortic flow (positive or negative diastolic flow) with none of change in its absolute value.

To find physiological reference data for the output impedance is very difficult. There is few data in literature. The application of our previous definition, that the output impedance of the vascular system can be described as the ratio of oscillatory pressure to oscillatory flow at the heart inlet valves, holds only for linear passive systems, or in other words for passive atria. There exists some difference between the output impedances for the model and the physiological data (Fig. 11). The physiological data are obtained by Fourier analysis of atrial outflow curves (i.e., ventricular filling curves) and atrial pressures, (curves from GREGG 1961), and the subsequent calculation of impedance spectra. The comparison shows that the physiological data differ mainly in the impedance moduli, and demonstrates simultaneously that the hydrodynamic inflow conditions for an artificial heart differ markedly from a natural heart, because of the lack of an active atrium.

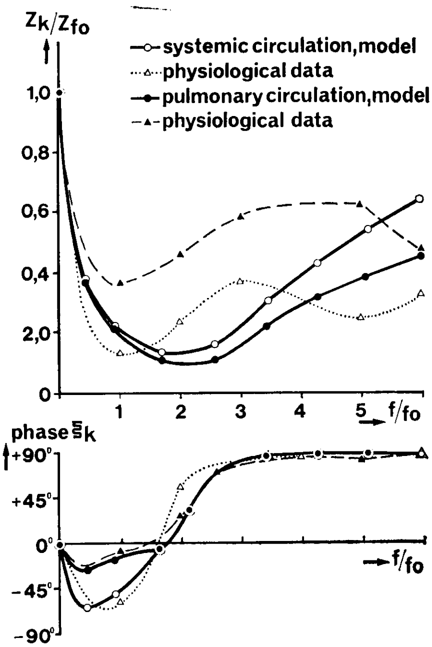

Fig. 11. Output impedance spectra of both the systemic and pulmonary circulations compared with physiological data.

4. Conclusion

Development of a lumped parameter hydraulic model has been carried out as a testing method for artificial hearts. This model is characterized by the input and output impedances of the circulatory system. Thus, the predominant goal was to mimic the physiological impedance spectra. Parameter estimation of the single model components has been conducted with an electrical analog model, which however incorporates already the unavoidable inertances of the hydraulic model. The inertances are introduced by the connecting tubes, by resistances which are not purely resistive, and by the four flow-probes at the inlets of the model system. As model data show, the desired input impedances are sufficiently good approximated and the input flow work rate spectra are in good conformity with the physiological one. This iterative optimization results in a compact integrated design of the hydraulic model of the systemic and pulmonary circulations.

Finally it should be pointed out that this model is especially designed to mimic the relevant impedance spectra and that it does not exihibit the phenomenon of wave travel. It is an extended Windkessel model and has no transmission-line characteristics.

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