

A hybrid (hydro-numerical) circulatory model: investigations of mechanical aortic valves and a numerical valve model

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Abstract. In most cases of diseased heart valves, they can be repaired or replaced with biological or mechanical prostheses. Biological prostheses seem to be safer than mechanical ones and are applied with good clinical outcomes. Their disadvantage, when compared with mechanical valves, is durability. In the development and application of mechanical and biological heart valves, a significant role can be played by a Hybrid (Hydro-Numerical) Circulatory Model. The aim of this paper is to demonstrate the opportunities created by the hybrid model for investigations of mechanical heart valves and their computer models under conditions similar to those of the circulatory system. A *diode-resistor* numerical valve model and three different design mechanical aortic valves were tested. To perform their investigations, computer applications were developed under RT LabView to be run on a PC. Static and dynamic characteristics of the valves were measured and registered - pressure in the numerical *time-varying elastance* left ventricle (p_{LV}), in the aorta ($p_{a,s}$) and flow (f), proving, among other factors, that 1) time delay of $p_{a,s}$ with respect to p_{LV} is mainly related to the valve's opening time, and 2) the valves of substantially different designs tested under identical hydrodynamic conditions reveal nearly the same dynamic performance.

Key words: biological and mechanical heart valves, numerical heart valve, hybrid (hydro-numerical) circulatory model, voltage-controlled hydraulic pressure (flow) source, hydraulic input impedance, *varying-elastance* left ventricle.

1. Introduction

Cardiac surgery is nowadays the method of choice for the treatment of diseased heart valves. In most cases, it offers better survival and a better quality of life when compared to nonsurgical methods [1, 2]. There are two options in heart valve surgery. The diseased valve can either be repaired or replaced. The repair procedure is often feasible in mitral valve surgery, especially in experienced hands. In reference centres, most mitral valves are spared [3]. Aortic valve sparing surgery is a more complex issue where the anatomy of the aortic root must be considered [4]. Only in a relatively small number of patients can such a procedure be performed. In other words, aortic valve replacement with either biological or mechanical prosthesis remains the gold standard; this inspires the search for a perfect valve. Also, such a valve, by definition, should be competent. There are two other criteria to consider when deciding upon potential perfection: ideal surface and ideal flow. The ideal surface means a biocompatible one which, in contact with blood, will not trigger the coagulation cascade. In terms of being smooth, the healthy endothelium lining human vessels is unrivalled by any artificial material. None of the mechanical valves currently produced have this feature, which means that patients with mechanical valves require permanent anticoagulation therapy and have to face the risk of thromboembolic events or bleeding. The bio-

logical (tissue) prostheses seem to be safer in this respect and are applied with good clinical outcome [5–7]. They do not require such an aggressive anticoagulation regime and allow for a more convenient life style. Their disadvantage, however, compared with mechanical valves, is durability. Patients with a biological valve must consider degeneration and undergo repeat surgery within 15-20 years. With biocompatibility on the biological side and durability on the mechanical side, it is the flow characteristics which can influence the final decision regarding the choice of valve. The term “ideal flow” means no gradient and no turbulence. The most common disease of aortic valve is aortic stenosis, which is usually caused by calcific degeneration. In aortic stenosis, the left ventricle must work against high impedance; this results in myocardial hypertrophy and finally, in cardiac failure, if not treated. If a new valve is implanted promptly, the recovery of the left ventricle may be initiated. This process is well described by left ventricle mass regression which simply means that it does not have to work as hard. Here, the better the parameters of a new valve, the higher the chance of a full recovery; this means a better survival in the long-term. For this reason, it is crucial that a good valve is in place. According to the American Heart Association (AHA) guidelines, there are three criteria when considering the severity of aortic stenosis: valve surface area, peak flow velocity and mean transvalvular gradient. It is quite

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a challenge to construct a valve allowing for laminar, with no gradient flow, reaching up to 10 l/min across the cross section area of 2–4 cm². Several studies referred to the mechanics and performance of artificial valves [8–12]. However, the debate is still ongoing. Any digital model that is able to match this situation would be welcomed by clinicians dreaming of a “perfect valve”.

In the development and application of mechanical and biological heart valves, a significant role can be played by a Hybrid (Hydro-Numerical) Circulatory Model (HCM) which was designed at the Institute of Biocybernetics and Biomedical Engineering (IBIB) of the Polish Academy of Sciences (PAN), in close co-operation with the Institute of Clinical Physiology (IFC), of the Italian National Council of Research (CNR) in Pisa, Section of Rome. The aim of this paper is to demonstrate the opportunities created by the hybrid model for investigations of mechanical heart valves and their computer models under conditions similar to those of the circulatory system where the ventricle is a *time-varying elastance* flow generator and ventricle-valve interactions are observed. The results of investigations on the HCM will contribute to the further development of heart valve surgery thanks to a better understanding of valve-circulation interactions.

2. Materials and methods

2.1. Mechanical heart valves.

Three mechanical heart valves with different constructions were selected for investigation under identical experimental conditions: Starr-Edwards caged-ball valve, the oldest one, implanted for the first time in 1960, consisting essentially of a silastic ball in a metal bearing cage [13]. To improve its performance, the valve underwent many modifications during the next few years, resulting in a caged-disc solution. However, due to the inferior hemodynamic features of caged-disc valves, such as flow turbulence, they are no longer implanted. It is remarkable, however, that clinicians still meet older patients in whom valves of this type were implanted many years ago and still have acceptable performance.

The Björk-Shiley tilting disc valve has been clinically applied since 1969 [14, 15]. The early model was modified in order to reduce the possibility of thrombus formation and it was approved for the market in 1979 [16]. For years, the Björk-Shiley heart valve was one of the most commonly implanted mechanical prostheses. However, its use was discontinued in 1986 after a report of a case of prosthesis malfunction due to disc dislocation [16]. However, the Björk-Shiley Delrin-Disc aortic valve earned respect as it was implanted for more than 30 years [15], with the number of procedures exceeding 360,000 in 2002 [17].

The St. Jude Medical mechanical bi-leaflet heart valve was introduced in 1979, and there have been more than 600,000 implantations since then [18]. It can be used in both aortic and mitral (atrio-ventricular) positions [18, 19]. At present, 80% of implanted mechanical valves are bi-leaflet ones, characterised by the lowest risk of thromboembolism [17].

2.2. Hybrid (hydro-numerical) model of the circulatory system: computer application for mechanical heart valves investigation.

The general layout of a computer application for mechanical heart valve testing in the hybrid (hydro-numerical) model of blood circulation is presented in Fig. 1 [20].

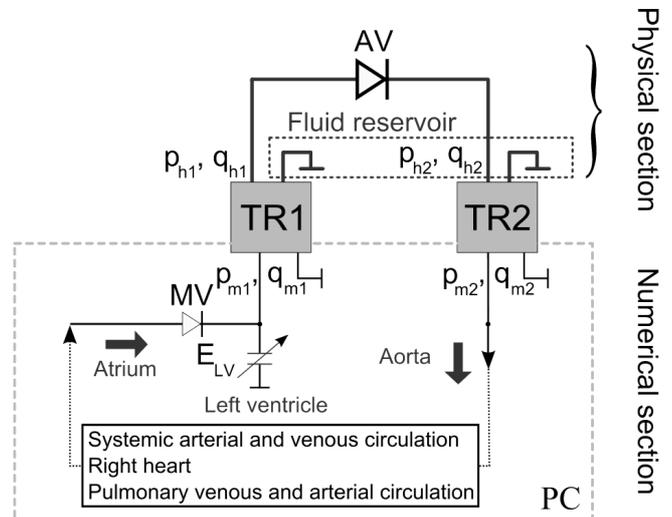


Fig. 1. General layout of the hybrid (hydro-numerical) model application for mechanical heart valves investigation: AV – mechanical aortic valve; MV – numerical mitral valve; TR1, TR2 – impedance transformers; E_{LV} – time-varying elastance of the numerical left ventricle; p_{m1} , q_{m1} and p_{m2} , q_{m2} – numerical pressures and flows, p_{h1} , q_{h1} and p_{h2} , q_{h2} – physical pressures and flows produced by impedance transformers TR1 and TR2, respectively

The mechanical aortic valve (AV) to be tested was inserted into the hybrid model by means of two impedance transformers (TR1 and TR2) [21], which were placed between the numerical and hydraulic sections. TR1 and TR2 convert numerical pressure and flow signals into their hydraulic equivalents and *vice versa*.

The functioning of transformer TR1 is based on a voltage-controlled hydraulic pressure source and TR2 on a voltage-controlled flow source. In both cases, the fluid reservoir of the model’s physical section was filled with a “working fluid”, which is a glycerol-water solution assuring a fluid viscosity of 3.6 Pa·s, approximately equal to the viscosity of blood.

It should be emphasised that in the model almost all liquid condensers were moved to the numerical section. This provided a step towards an ideal model structure where, for all elements of the system during experiments, unlimited changes of parameters were possible via the user interface.

The voltage-controlled hydraulic pressure (flow) source is presented in Fig. 2. In both cases (pressure and flow), this was based on the gear pump GP inserted between the incompressible working fluid container C_f and the pneumo-hydraulic small volume condenser C_h stabilising the system. A driving system of the pump consisted of an electrical DC motor M (Maxon RE65), a tachogenerator T (Hübner GTL5) and a controller CON 1 (Maxon ADS 50/10) to form together the hybrid pressure (flow) source. Two converters, p/u and AD,

produced a digital pressure signal p and delivered it to a real-time computer PC (National Instruments) run on LabVIEW RT. The signal was filtered in a low pass filter F to remove high frequency noise, and then was sent to controller CON 2 to be compared to a set value of pressure p_0 . The output signal of the controller was transmitted to an input of a program switch PS.

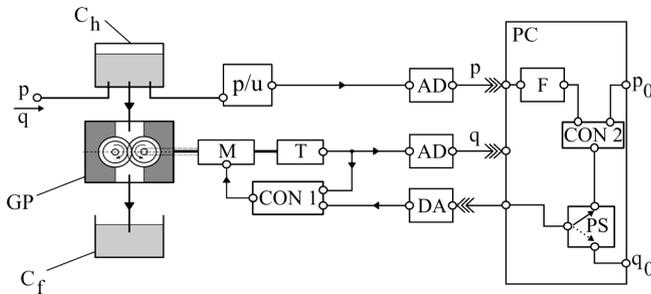


Fig. 2. Hybrid pressure/flow source: p – pressure; q – flow; C_h pneumo-hydraulic condenser; C_f – “working” fluid container; M – electrical motor; T – tachogenerator; CON 1, 2 – controller 1, 2; AD, DA – analogue-to-digital and digital-to-analogue converter; F – low pass filter; p_0 (q_0) – input pressure (flow); PS – program switch; p/u – pressure to voltage converter; GP – gear pump; PC – personal computer

When required, the system can also work as a flow source. In that case, the input value of the controller 2 was flow q_0 .

The pressure (flow) sources described allowed the design of basic functional modules i.e. hybrid interfaces merging numerical (computer) and physical (hydraulic) signal environments simultaneously playing a role of impedance transformers (TR), presented in detail in [22] and in the Appendix to this paper. Thus, it is necessary to consider an exemplary scheme in Fig. 3 where the impedance transformer TR consists of the voltage-controlled flow source VCFS enabling q_0 input, analogue-to-digital (AD) and digital-to-analogue (DA) converters and a pressure sensor p/u . When the input pressure p_h was delivered to the numerical input impedance Z_m of the circulatory model as the input numerical pressure signal, p_m , this evoked an input numerical flow q_m . Flow q_m was then converted into analogue set flow q_0 . As a result, VCFS produced an input flow q_h in the hydraulic input of TR.

Finally, the hydraulic input impedance Z_h is directly proportional to the input impedance Z_m of the model, which can be written as:

$$Z_h = K \cdot Z_m, \quad (1)$$

for $K = 1$:

$$Z_h = Z_m. \quad (2)$$

This means that the numerical solution (variables p_m , q_m) of a set of both differential or/and algebraic equations representing the circulatory model can be transferred to the hydraulic input of TR in the form of pressure p_h and flow q_h . One TR can be connected to an arbitrarily selected point of the circulatory model, e.g. the hydraulic input connected to open-fluid container C_f by a hydraulic resistor; “the patient” represented by the circulatory model in that case will be

“bleeding” and losing the “numerical” blood converted into “physical” blood flowing through the physical resistor.

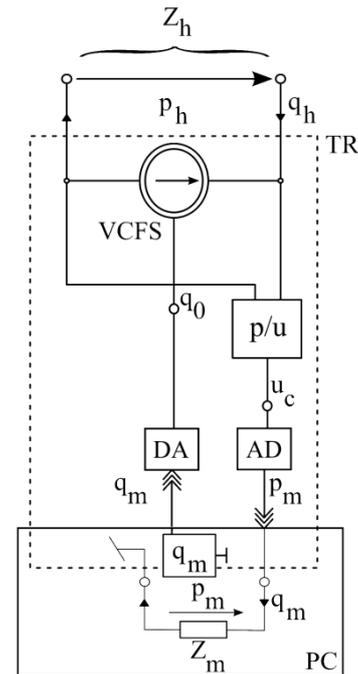


Fig. 3. Impedance transformer (TR): VCFS - voltage-controlled flow source; q_0 – analogue set input flow; AD, DA - analogue-to-digital and digital-to-analogue converters; p/u – pressure sensor; p_h , q_h – hydraulic pressure and flow; p_m , q_m – numerical pressure and flow; Z_h , Z_m – hydraulic and numerical input impedance

2.3. Computer application for artificial valves static characteristics. To determine the static characteristics of mechanical aortic valves, a computer application was developed to be run on real-time PC. A simple layout of the application is presented in Fig. 4.

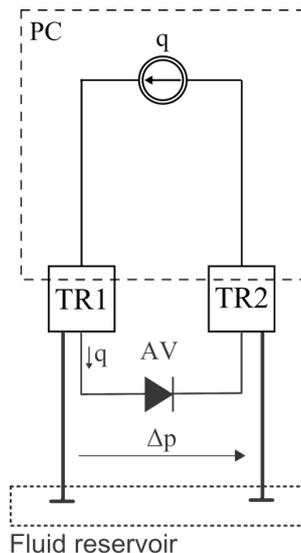


Fig. 4. Computer application for mechanical valves static characteristics: AV – mechanical aortic valve; TR1, TR2 – impedance transformers; q – numerical flow; Δp – pressure drop on valve AV

In the numerical part of the model, a source of flow q was realised as a program section. That flow, thanks to transformers TR1 and TR2, could be converted from the numerical to physical form and passed through the mechanical valve to transformer TR2; in the figure, the valve is presented as a diode. To avoid inertial effects, slow variation of flow q was applied, so that the corresponding pressure drop, Δp , on valve AV could be measured and static characteristics $\Delta p(q)$ could be found.

3. Mechanical and numerical heart valves testing

3.1. Static characteristics. Static characteristics of the selected mechanical valves (Starr Edwards, St. Jude and Björk-Shiley) and of a *diode-resistor* numerical valve model (Fig. 5) are presented in Fig. 6. The characteristics of the numerical valve were obtained from measurements of two resistance values of *forward flow* direction R_{avf} : 30 and 15 $\text{g}\cdot\text{cm}^{-4}\cdot\text{s}^{-1}$. Mean values of the *forward flow* resistance of the mechanical valves were: $R_{\text{Starr-Ed}} = 93.4$; $R_{\text{St.Jude}} = 89.5$ and $R_{\text{Bjork-Shiley}} = 75 \text{ g}\cdot\text{cm}^{-4}\cdot\text{s}^{-1}$ when *back flow* resistance (R_{avb}) was constant and equal to $10^5 \text{ g}\cdot\text{cm}^{-4}\cdot\text{s}^{-1}$.

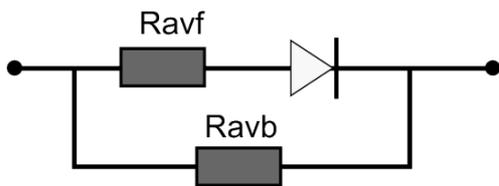


Fig. 5. A *diode-resistor* numerical valve: AV – valve; R_{avf} – valve *forward flow* resistance, R_{avb} – *back flow* resistance

In general, the static characteristics $\Delta p(q)$ of the mechanical valves are similar in shape and well approximated by second order polynomial:

- Starr-Edwards:

$$y = 0.0001x^2 + 0.0026x - 0.0419; \quad R^2 = 0.9994, \quad (3)$$

- Saint-Jude:

$$y = 1E - 04x^2 + 0.0011x - 0.0034; \quad R^2 = 0.9997, \quad (4)$$

- Björk-Shiley:

$$y = 7E - 05x^2 + 0.0036x + 0.0308; \quad R^2 = 1, \quad (5)$$

where $E = 10$, and R is the correlation factor.

In the same diagram (Fig. 6), the static characteristics of the *diode-resistor* valve show imperfection of the numerical linear model wrongly reproducing static valve properties for flows exceeding 200–300 $\text{cm}^3\cdot\text{s}^{-1}$ (12*-18 l/min). It is necessary to remember that even if typical cardiac output $\text{CO} = 6 \text{ l/min}$ in physiological heart conditions, in *systole* the instantaneous flow through the aortic valve can exceed 500 cm^3/s (30 l/min).

Nevertheless, the obtained results allow the assumption that the computer application functioned properly.

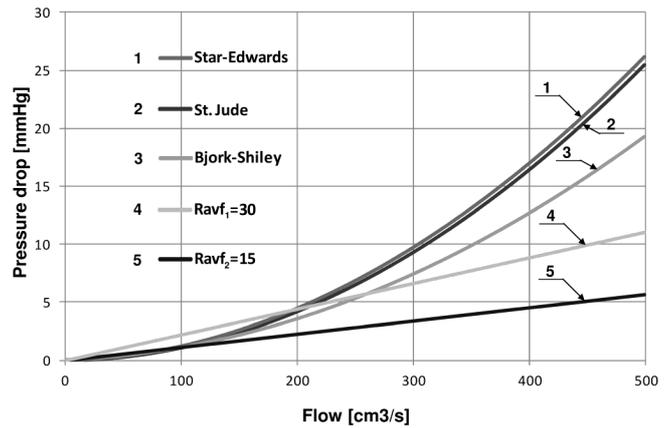


Fig. 6. Static characteristics of mechanical and numerical aortic valves: 1 – Starr Edwards; 2 – St. Jude; 3 – Björk-Shiley; 4 – $R_{avf} = 30 \text{ g}\cdot\text{cm}^{-4}\cdot\text{s}^{-1}$; 5 – $R_{avf} = 15 \text{ g}\cdot\text{cm}^{-4}\cdot\text{s}^{-1}$

3.2. Dynamic characteristics. In the case of each tested aortic valve the time courses of three hemodynamic parameters were measured: pressure in numerical *time-varying elastance* left ventricle (p_{LV}) and in the aorta (p_{as}) as well as flow in the aorta (f). The courses presented in Figs. 7–10 were registered for *forward flow* direction valve resistance $R_{avf} = 30 \text{ g}\cdot\text{cm}^{-4}\cdot\text{s}^{-1}$.

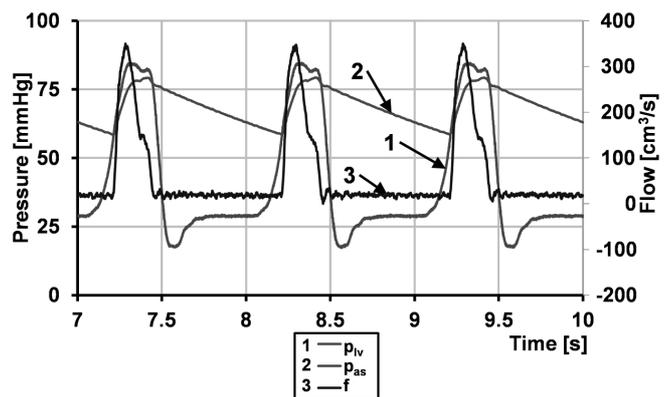


Fig. 7. *Diode-resistor* numerical model aortic valve time courses: 1 – left ventricular pressure (p_{LV}), 2 – aortic pressure (p_{as}), 3 – aortic flow (f)

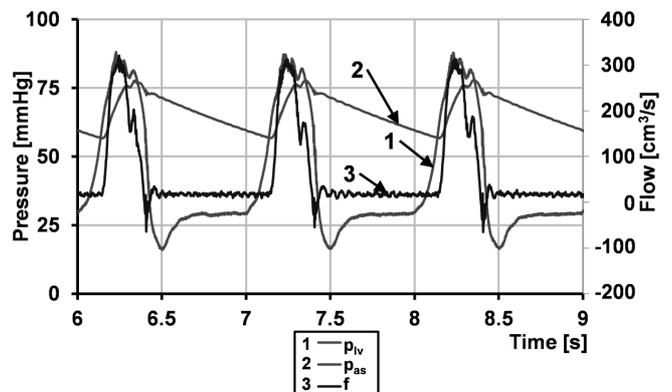


Fig. 8. Björk-Shiley aortic valve time courses: 1 – left ventricular pressure (p_{LV}), 2 – aortic pressure (p_{as}), 3 – aortic flow (f)

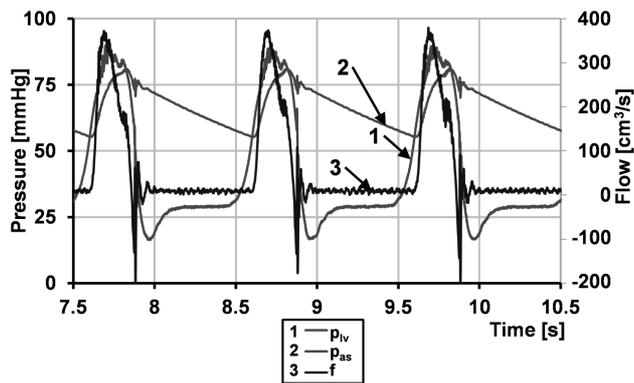


Fig. 9. St. Jude aortic valve time courses: 1 – left ventricular pressure (p_{LV}), 2 – aortic pressure (p_{as}), 3 – aortic flow (f)

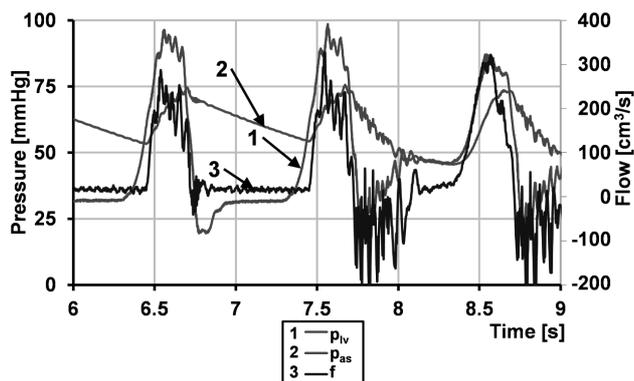


Fig. 10. Starr Edwards aortic valve time courses: 1 – left ventricular pressure (p_{LV}), 2 – aortic pressure (p_{as}), 3 – aortic flow (f)

All of the investigations were conducted using the same pathological circulatory conditions of the hybrid model, with left ventricular elastance $E_{LV} = 1 \text{ mmHg}\cdot\text{cm}^{-3}$ and rest volume $V_{ol} = 15 \text{ cm}^3$, and systemic arterial resistance $R_{as} = 1316 \text{ g}\cdot\text{cm}^{-4}\cdot\text{s}^{-1}$ ($1 \text{ mmHg}\cdot\text{cm}^{-3}\cdot\text{s}$).

4. Results

A computer application designed to determine static and dynamic characteristics of artificial heart valves allowed tests of a numerical and various mechanical heart valves to be performed. The results are as follows:

1) Time courses of the mechanical valves proved time delays of the aortic pressures (p_{as}) in relation to the corresponding left ventricular pressures (p_{LV}) (Figs. 8–10), which are mainly related to the opening times of the valves: the longer the opening time, the bigger the aortic pressure delay. In the case of the numerical *diode-resistor* valve model, a zero valve opening time was followed by zero aortic pressure delay, which can be clearly seen in Fig. 7. The aortic pressure delays are consistent with the corresponding opening times of the tested mechanical valves (Figs. 8–10).

Beneath, the opening times of the mechanical and numerical heart valves are presented:

- St. Jude – 25 ms;
- Starr Edwards and Björk-Shiley – 30 ms;
- numerical *diode-resistor* valve model – 0 ms.

2) As far as closing times are concerned, valve closing times significantly longer than zero caused the back flow and dicrotic notch to appear on the aortic pressure courses, which is characteristic for mechanical valves (Figs. 8–10). Contrary to that, prompt closing of the numerical valve, with no swings and vibrations, resulted in no back flow and no dicrotic notch.

3) The tested valves were characterised by different valve resistance; this also resulted in different values of the same hemodynamic parameters in identical experimental conditions. Moreover, rising valve resistance was followed by an increase in the ventricular – aortic pressure difference and finally by an adverse oxygen supply loss. Ventricular – aortic pressure difference of the mechanical valves tested in the same hemodynamic circulatory conditions is clearly evidenced in the set of values presented in Table 1.

Parameter → valve ↓	R_{avf} [$\text{g}\cdot\text{s}^{-1}\cdot\text{cm}^{-4}$]	$p_{LV\max}$ [mmHg]	$p_{as\max}$ [mmHg]	$p_{LV\max} - p_{as\max}$ [mmHg]
Starr-Edwards	93.4	92	75	17
St. Jude	90	93	81	12
Björk-Shiley	75	86	77	9
Numerical valve	30	84	79	5

R_{avf} – valvular forward flow resistance,

$p_{LV\max}$, $p_{as\max}$ – left ventricular and aortic pressures.

4) During the experiments, some problems with the Starr-Edwards valve arose in *diastole* related to the valve closing; this can be clearly seen when two different-shape heart cycles are compared (Fig. 10). As shown by further experiments, impaired closing of the valve was the result of occasional improper and unstable positioning of the valvular ball, possibly caused by the aortic pressure being too low at that moment.

5. Conclusions

The investigations confirmed proper and stable functioning of the computer application developed to adapt the hybrid circulatory model to test mechanical heart valves of substantially different designs. During those tests, important hydro-mechanical and hemodynamic parameters were measured and registered to evaluate static and dynamic features of the tested valves, including valve resistance affecting both forward and reverse flow, as well as valve opening and closing times. There was a remarkable difference between numerical valves and mechanical ones regarding the presence of dicrotic notch and back flow. These can be observed in mechanical valves because of the relatively long opening and closing times (tens of milliseconds). Despite some evident drawbacks in the numerical valve model applied, its performance is quite promising in terms of further study.

It is also worth pointing out that the dynamic properties of the tested valves were practically the same in spite of quite different valve designs, which could allow the authors to conclude that the opening and closing times mainly depended on valvular-ventricular-aortic interactions.

6. Summary

Cardiac surgery is one method for the treatment of diseased heart valves. In most cases, it offers better survival and better quality of life when compared to nonsurgical methods [1, 2]. The diseased valve can either be repaired (often feasible in mitral valve surgery) or replaced. Those patients with mechanical valves require permanent anticoagulation therapy and have to face the risk of thromboembolic events or bleeding. Biological prostheses seem to be safer in that respect and are applied with good clinical outcome [5–7]. Their disadvantage, when compared with mechanical valves, is the smaller durability. A patient with a biological valve must consider its degeneration and a follow-up surgery in 15–20 years. In the development and application of mechanical and biological heart valves, a significant role can be played by a Hybrid (Hydro-Numerical) Circulatory Model which was designed at the Institute of Biocybernetics and Biomedical Engineering (IBIB) of the Polish Academy of Sciences (PAN) in close co-operation with the Institute of Clinical Physiology (IFC), of the Italian National Council of Research (CNR) in Pisa, Section of Rome. The aim of this paper was to demonstrate the opportunities created by the hybrid model for the investigation of mechanical heart valves and their computer models under conditions similar to those seen in the circulatory system. A *diode-resistor* numerical valve model and three different mechanical aortic valves were selected to the experiments: the Starr-Edwards caged-ball valve, the Björk-Shiley tilting disc valve and the St. Jude Medical mechanical bi-leaflet valve. To perform these investigations, computer applications were developed under RT Lab-View to be run on a PC. Static characteristics and time courses of three haemodynamic variables were registered: pressure in the numerical *varying-elasticance* left ventricle (p_{LV}) and in

the aorta (p_{as}) and flow in the aorta (f). The results showed the following: 1) the opening times of the tested heart valves were: for the numerical *diode-resistor* valve model – 0 ms, for the Starr Edwards and Björk-Shiley – 30 ms and for St. Jude – 25 ms. In addition, time delays of aortic pressures p_{as} in relation to left ventricular pressures p_{LV} were different – they were smaller for shorter and larger for longer opening times (Figs. 8–10); 2) the tested aortic valves are characterised by different valvular resistance; this results in different values for the same haemodynamic parameter under identical experimental conditions. Moreover, rising valve resistance is followed by increasing ventricular and aortic pressure differences and finally, by adverse oxygen supply losses. Ventricular-aortic pressure differences of the tested mechanical valves in identical circulatory conditions are clearly shown in Table 1. 3) During the experiments, some problems with Starr-Edwards valves arose in *diastole* related to valve closing – this can be clearly seen when two different-shape heart cycles are compared (Fig. 10). As further experiments showed, impaired closing of the valve is a result of the occasional improper and unstable position of the valve ball, possibly caused by too low aortic pressure.

The investigations confirmed proper and stable functioning of the computer application developed to adapt the hybrid circulatory model to test mechanical heart valves of substantially different designs.

Appendix

In Fig. A1 a block diagram of a whole closed-loop hybrid (hydro – numerical) circulatory model is presented and its connection to the mechanical heart valve (AV). The valve is

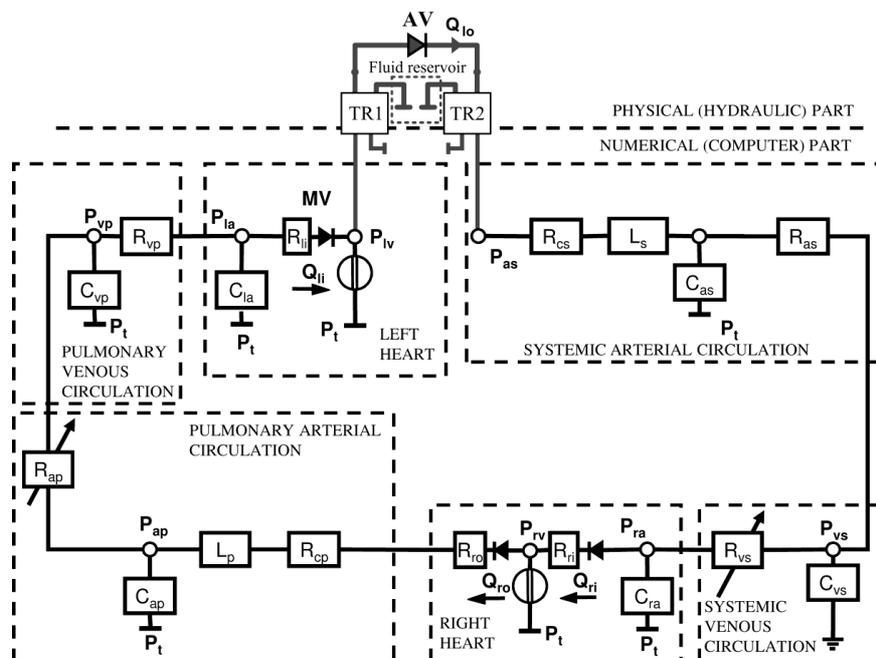


Fig. A1. Block diagram of the closed-loop hybrid (hydro – numerical) circulatory model and its connection to the mechanical aortic heart valve (AV)

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merged into the circulatory model in aortic position. The liquid running from the “numerical left ventricle” of the model has the nature of a “numerical flow” transformed by the impedance transformer TR1 into the hydraulic flow delivered from the fluid reservoir to the mechanical heart valve (AV) and then to the impedance transformer TR2, where it is again converted into the “numerical flow” to enable its direction to the numerical aorta and systemic arterial system while its previous physical liquid flows back to the reservoir.

List of symbols from Fig. A1 is presented in Table A1.

In the circulatory modeling the expression “numerical” is interchangeable with “computational” which is always electronic. Referring to a hybrid model itself, it is popular to realize numerical section by means of RLC elements (resistors, coils and condensers) to mimic resistive, inertial and capacitive features of simulated objects and to create in this way a numerical (computational) model built of electrical components.

Table A1
List of symbols

Abbreviations	Variables
TR1	Numerical/Physical (hydraulic) Impedance Transformer
TR2	Physical/Numerical Impedance Transformer
AV	Mechanical Aortic Valve
MV	Numerical mitral valve
L_s/L_p	Systemic/Pulmonary Inertance
C_{as}/C_{ap}	Systemic/Pulmonary compliance
R_{cs}/R_{cp}	Systemic/Pulmonary Characteristic Resistance
R_{as}/R_{ap}	Systemic/Pulmonary Arterial Resistance
P_{lv}	Left Ventricular Pressure
p_{as}	Systemic Arterial Pressure
p_{la}	Left Atrial Pressure
P_{rv}	Right Ventricular Pressure
p_{ap}	Pulmonary Arterial Pressure
p_{ra}	Right Atrial Pressure
C_{la}/C_{ra}	Left/Right Atrial Compliance
C_{vs}/C_{vp}	Venous Systemic/Venous Pulmonary Compliance
R_{ti}	Mitral Valve Resistance
Q_{ti}	Mitral Valve Flow
Q_{to}	Mechanical Aortic Valve Flow
R_{ri}	Tricuspid Valve Resistance
R_{ro}	Pulmonary Valve Resistance
Q_{ri}	Tricuspid Valve Flow
Q_{ro}	Pulmonary Valve Flow
P_{vs}	Venous Systemic Pressure
R_{vs}	Venous Systemic Resistance
P_{vp}	Venous Pulmonary Pressure
R_{vp}	Venous Pulmonary Resistance
P_t	Intrathoracic Pressure

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