Durability/Wear Testing of Heart Valve Substitutes

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Background and aims of the study: **The current standards for accelerated heart valve testing have considerable differences in test conditions. Another problem arises from the fact that such test systems are not standardized at all. It was shown earlier that different test systems generate totally different valve loading, even if operating at standard conditions. The present study aimed to improve this unsatisfactory situation and to develop a new concept where actual loading of valves is measured either in vitro or in vivo under physiologic conditions and subsequently to reproduce these conditions during accelerated testing.**
Methods: **Integral loading forces at valve closure were measured for several valve types using a piezoelectric force ring within a real-time circulatory mock loop under physiologic conditions. This facilitated definition of a physiologic loading range. Physiologic loading was subsequently reproduced in a single-chamber accelerated test system. Working conditions obtained in terms of stroke, bypass flow and compliance served as design criteria for a new test chamber and a complete 12-chamber accelerated testing system.**
Results: **The integral loading obtained using the force ring showed a correlation with previous in vitro and in vivo results of strain-gauged valves. Loading forces for mechanical valves are about one order of magnitude higher than for bioprosthetic valves and are strongly related to cardiac output for both valve types. At physiologic loading, however, the differential pressures across the valves are considerably below those given in FDA guidelines.**
Conclusions: **This pilot study demonstrates that physiologic valve loading is reproducible over a wide range under appropriate testing conditions. It also showed that, at the back-pressures of the current standards, the loading forces during accelerated testing exceed the real-time loading forces by far and, thus, may provide unrealistically high valve loads. These initial findings indicate that amendments of the currently valid standards may be need to be accorded.**

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In 1996 we published a critical review on the state of the art of heart valve wear and fatigue testing as represented by the current standards of ISO, FDA and CEN (1). These standards have considerable differences in test conditions as far as pressure differences across the test valves and total cycle numbers are concerned. Moreover, since the testing devices themselves are not standardized, different devices generate totally different loading conditions on the valves, even if the tests are carried out under the same standard. This study clearly demonstrated that, under the current standards, the conditions of actual, in vivo impact loading of a valve cannot be reproduced. Also, the two test devices which were compared, although both operating at the given standards conditions, generate totally different loading conditions. For one tester loading decreases with increasing cycle rate, while for the other it increases. For all testers, time history of loading and pressure difference across the valve are a function of test frequency, compliance and other factors.

To improve this unsatisfactory situation it was strongly suggested that the actual loading conditions of each valve type be measured either in animal models in vivo or within a circulatory mock loop which provides physiologic loading conditions; this loading could then be subsequently reproduced during accelerated wear testing. The present study is set out to investigate this concept.

Measurements of valve loading under physiologic conditions

Since most valves - and especially bioprostheses - are not suitable for the attachment of strain gauges due to the lack of appropriate fixation points which are representative for valve loading, a piezoelectric force measurement ring (Kistler, type 906 1A, Winterthur, Switzerland) (Fig. 1, top right) was used to assess loading at valve closure. This type of force transducer measures the integral loading force acting on the valve.

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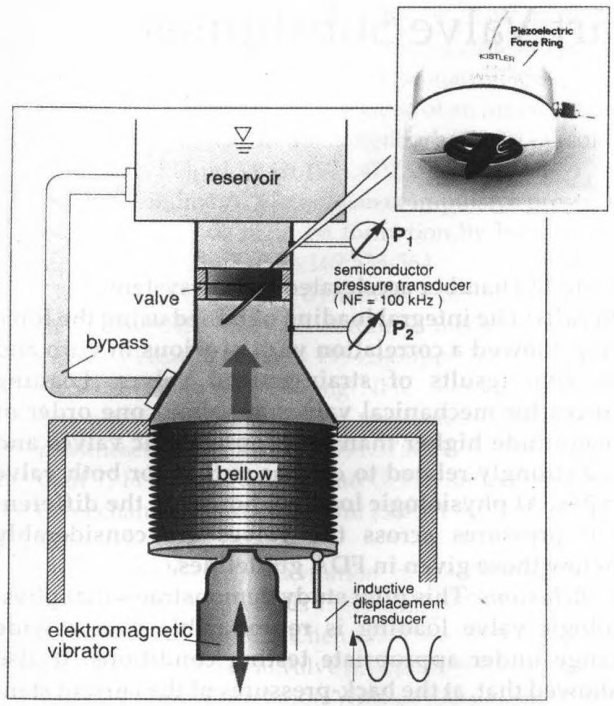

Figure 1: Schematic of single-chamber accelerated valve test system with piezoelectric force ring in the upper right.

The next step was to correlate these integral loading forces with previously obtained in vivo and in vitro results (2,3) measured by a strain-gauged 29 mm Björk-Shiley Convexo-Concave (BSCC) mechanical tilting disk valve. For this purpose, the strain-gauged BSCC valve was mounted on the force ring and inserted into the mitral position of a physiologic circulatory mock loop as previously described (4). Both, strut loads and integral loading force were measured simultaneously.

For these measurements the left ventricular dp/dt (according to the FDA guidelines averaged over the last 20 ms before valve closure) of the model ventricle was varied between 500 and 1500 mmHg/s by increasing the stroke volume at a fixed test rate of 70 per min, a mean aortic pressure of 100 mmHg, and a systolic duration of 15%.

The results are shown in Figure 2. The upper curve represents the force ring measurements which range from 25 to 110 N integral loading force at the corresponding dp/dt values. The two lower curves represent the strut loading forces obtained by strain gauges in vivo and in vitro for the same valve; these range from 2 to 20 N strut loading force. The difference in measured forces between the two methods is obvious and is related to the fact that the strain gauges measure only loading of a single strut caused by strut deflection, while the force ring measures the integral total loading force acting on the closed valve. Nevertheless, both

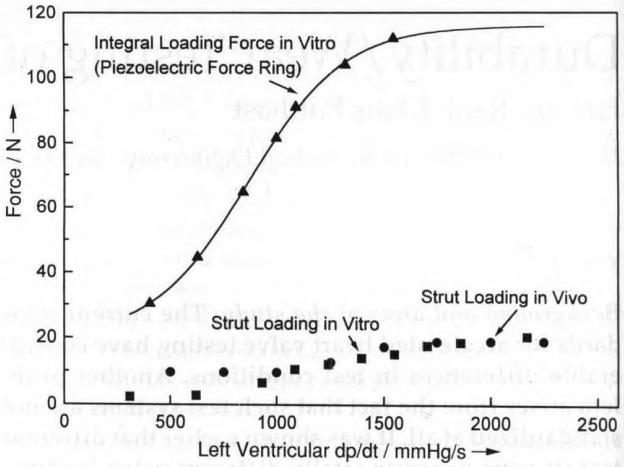

Figure 2: In vivo and in vitro loading of a 29 mm BSCC valve as a function of left ventricular dp/dt before valve closure.

methods give a good correlation with left ventricular dp/dt and, thus, can as well be correlated among each other.

Measurements under accelerated testing conditions

After establishing the above correlation between integral forces and left ventricular dp/dt the valve was inserted into a specifically developed single-chamber system for accelerated valve testing (also shown in Fig. 1).

An electromagnetic vibrator (Ling Dynamics, type 409, Royston, UK) generates a sinusoidal flow through the test valve by compression and extension of a metallic bellows. When the valve is closed, fluid from the lower chamber flows to the upper chamber via an adjustable bypass. This bypass serves simultaneously to control the pressure difference across the valve, which is measured upstream and downstream of the valve by two semiconductor pressure transducers with a natural frequency of 100 kHz (Cobe Disposable Transducer, Lakewood, CO, USA). Test fluid is water at room temperature.

The test rate was increased from 200 to 2000 per min and the integral loading forces were measured. The pressure difference across the valve was kept constant at 120 mmHg for all test rates. The results are presented in Figure 3.

The shaded area represents the range of loading forces under all potential physiologic conditions as obtained in vivo and in vitro and as depicted in Figure 2. The central curve shows the measured integral loading forces under accelerated testing conditions under observance of the FDA conditions for accelerated valve testing such as full opening and closing and 120 mmHg pressure difference. The minimally adjustable loads

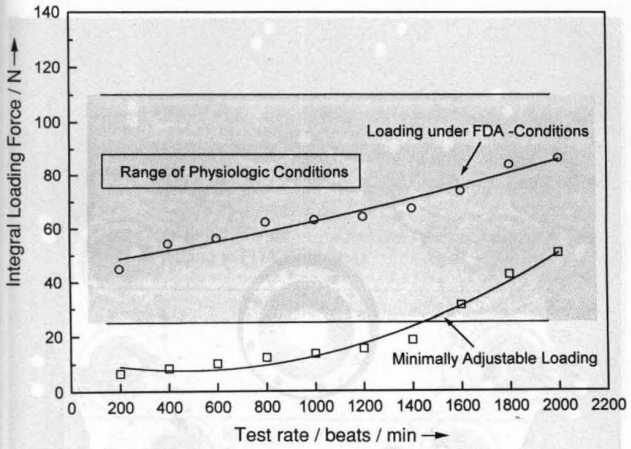

Figure 3: Integral loading forces within the single chamber accelerated test system obtained for a 29 mm BSCC valve at a back-pressure of 120 mmHg (FDA-condition).

are represented by the lower curve. Higher loads can be easily generated by adjusting stroke and bypass throttle of the test system. These results clearly show that physiologic valve loading can be reproduced over a wide range under appropriate accelerated testing conditions. It is also evident that, under FDA conditions, only a single loading force which increases with higher test rates can be generated for a defined test rate, whereas test conditions outside of the current standard in terms of pressure difference facilitate the adjustment of a range of physiologic loading conditions at any test rate. Thus, valve loading at resting conditions as well at exercise conditions can be adjusted, resulting in a much better simulation of the loading history of an implanted valve.

Valve loading forces for alternative valve types

So far, all presented results are only valid for one single valve type and size, a 29 mm diameter BSCC tilting disk valve. Therefore, in order to develop a generally valid concept, it is necessary to extend the above findings to other valve types and sizes. For this purpose, a St. Jude Medical (SJM) 27 mm mechanical bileaflet valve and an Ionescu-Shiley 25 mm pericardial bioprostheses were selected as a first step. Both valves were inserted into mitral position of our circulatory mock loop and integral loading forces were measured under the following experimental conditions (according to FDA guidelines for pulsatile flow valve testing):

Cardiac output: 2 l/min and 7 l/min;
Test rate: 70 per min;
Mean aortic pressure: 100 mmHg;
Mean atrial pressure: 10 mmHg; and
Systolic duration: 35%.

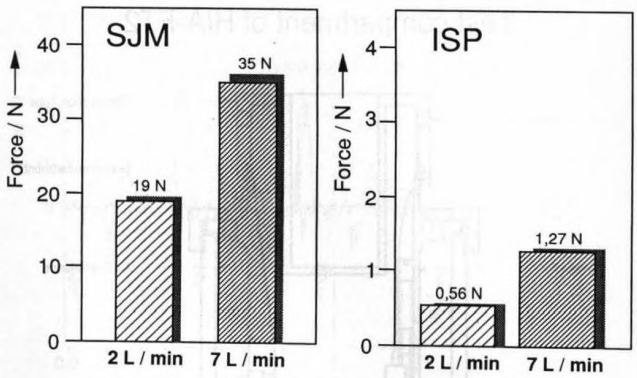

Figure 4: Integral loading forces for a 27 mm SJM mechanical valve and a 25 mm ISP bioprosthesis measured under real-time conditions within a circulatory mock loop.

This range of test conditions represents the normal physiologic range and can be used for the characterization of corresponding loading conditions. For each valve type the measured loading forces were averaged over 100 cycles.

The results (Fig. 4) show that loading for the mechanical valve is about one order of magnitude higher than for the bioprosthetic valve, and loading increases with a factor of about two when cardiac output changes from 2 l/min to 7 l/min.

Design of a new fatigue tester

The principal design criteria for a valve test chamber in terms of stroke, displaced volume, bypass flow and compliance were obtained by means of the above-mentioned single-chamber system (see Fig. 1). Based on these criteria a new test compartment for general use was designed (shown schematically in Fig. 5). For the new design, the electromagnetic vibrator was replaced by a swash plate with adjustable stroke. Otherwise, the operating principle was the same as already described above.

Larger valves generate higher impact loading forces than smaller ones. Therefore, since the new test compartment is intended for use with all kinds of valve sizes, an adjustable compliance chamber was added for additional control of loading forces. The desired impact force can then be adjusted by varying the following four parameters:

- stroke of the swash plate;
- test rate;
- air compliance; and
- bypass throttle adjustment.

The FDA guidelines require that for all valves, testing should be conducted on three of the largest, medium and smallest of each valve type. One equivalent tis-

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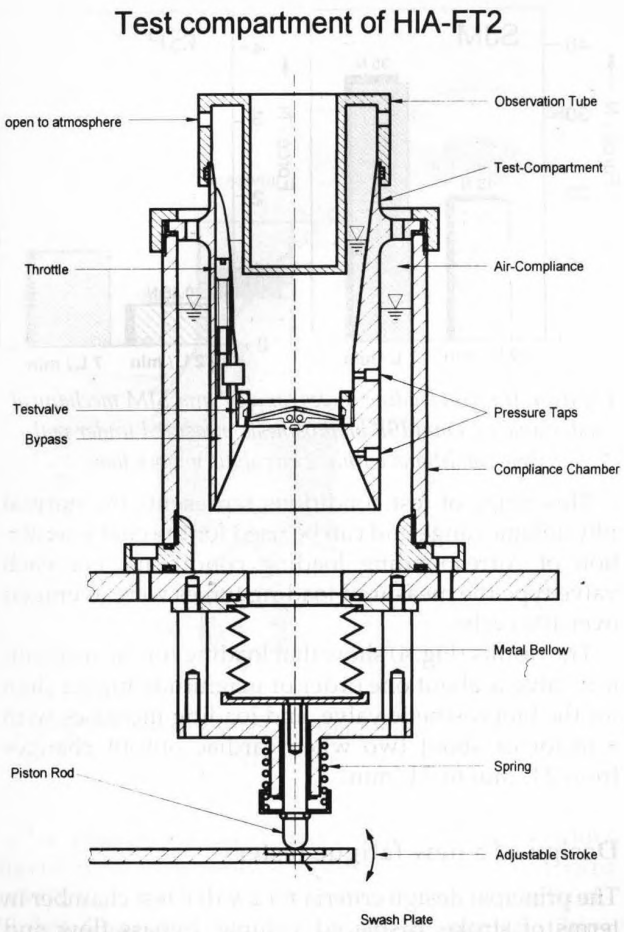

Figure 5: Schematic cross-section of newly developed test compartment for accelerated valve testing.

sue annulus diameter of each type reference valve must be tested under identical conditions. This results in a total of 12 test valves. Accordingly, a 12-cylinder fatigue tester (HIA-FT2) has been designed and manufactured (Fig. 6).

The first two of the above-listed parameters cannot be adjusted individually; they are the same for all 12 test compartments. Thus, after the correct adjustment of one compartment the other 11 must be adjusted by variation of air compliance and bypass throttle flow. An example (Fig. 7) shows where the peak loading forces for a 27 mm SJM valve have been varied by changing the air compliance at a constant test rate, bypass throttle and stroke settings. Peak load decreases linearly with increasing compliance volume and can easily be adjusted for physiologic loading conditions.

Verification of concept within new fatigue tester

In a final step a 27 mm mechanical SJM valve and a 25 mm Ionescu-Shiley pericardial valve (ISP) were inserted

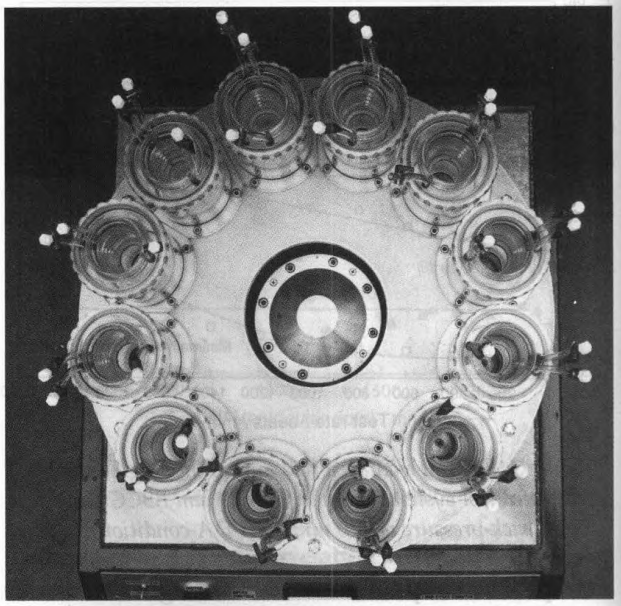

Figure 6: Top view of newly developed 12-chamber accelerated test system (HIA-FT2).

into the new test compartment and the loading forces together with the pressure difference across the valves were measured within the fatigue tester at a test rate of 1000 per min for the SJM and 600 per min for the ISP, respectively. The results (Figs. 8 and 9) show that for both valve types the peak loading forces are within the previously determined physiologic range (compare Fig. 4). The differential pressures, however, are considerably below the pressures given in the FDA guidelines, which should be adjusted to 120 mmHg for both valve types.

An example illustrating this controversy for the SJM valve (Fig. 10) shows that, in this case, the pressure difference across the valve was adjusted to 120 mmHg by reducing the test chamber compliance. Test rate, bypass flow and stroke were kept constant. As can be clearly seen, valve loading forces exceed 50 N and, thus, are far above physiologic loading.

Discussion

This pilot study demonstrates that physiologic valve loading can be reproduced under appropriate accelerated testing conditions. However, testing under these conditions requires special testing compartments with an increased number of control parameters for the adjustment of physiologic loads. It also requires a two-step testing approach: first, physiologic loading forces have to be determined within a real-time circulatory mock loop; and second, this loading has to be reproduced within the accelerated tester. For proper transfer of physiologic real-time loading forces to accelerated testing the test valves have to be mounted within a cal-

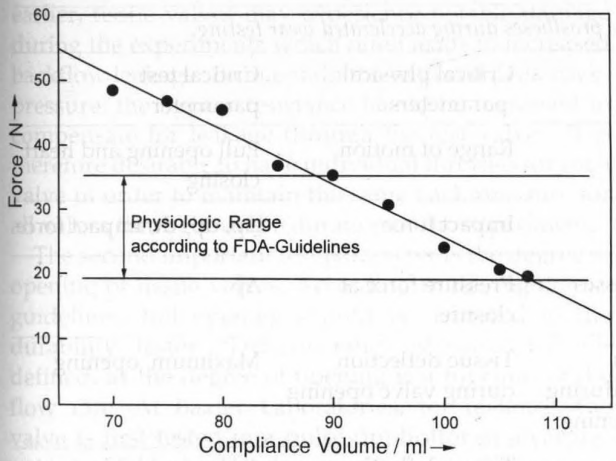

Figure 7: Integral loading force as a function of air compliance volume measured for a 27 mm SJM valve within the new test compartment at constant test rate (1000 per min), bypass throttle and stroke settings.

ibration compartment, equipped with a force ring transducer. Once physiologic loading is adjusted and the parameter settings in terms of stroke volume, compliance volume and bypass flow are obtained, the valves are mounted within the geometrically similar final test compartments and the continuously measured pressure difference serves as the single control parameter for long-term studies.

It could also be shown that at the back-pressures of the current standards the loading forces during accelerated testing exceed the real-time loading forces by far and thus may provide unrealistically high valve loads.

These initial findings indicate that corresponding amendments of the currently valid standards may be necessary.

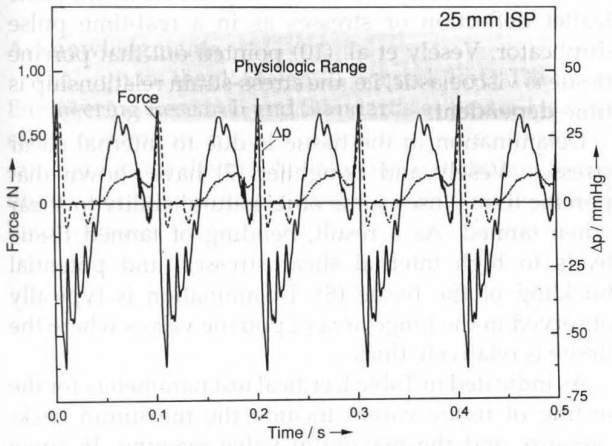

Figure 9: Integral loading force and pressure difference across valve for a 25 mm ISP bioprosthesis at a test rate of 600 per min.

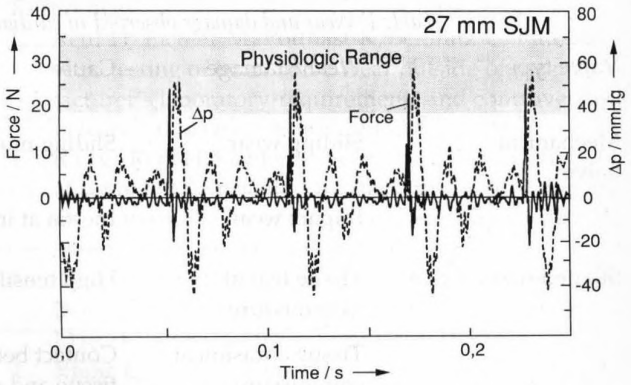

Figure 8: Integral loading force and pressure difference across valve for a 27 mm SJM valve at a test rate of 1000 per min.

Durability testing of bioprosthetic heart valves requires some additional discussion. The FDA and ISO guidelines for the wear testing of tissue valves are very similar to those for mechanical valves. Valves are to be tested for an equivalent of five years at a peak back-pressure of at least 90 mmHg for aortic and 120 mmHg for mitral valves. The valves should open and close completely. For stentless valves, the aortic wall should be modeled by a compliant tube.

Unlike mechanical valves, tissue valves may show severe damage as a result of the wear test and may become dysfunctional before completion of the experiment. Possible damage to tissue valves include tissue tear, holes, delamination, abrasion, prolapse and stent fracture (5,8,9). Table I lists commonly observed damage in tissue valves, their cause, and critical test parameters that influence the damage. Wear in mechanical valves is also included as a reference.

Tissue tear is typically caused by high tensile stresses. Finite element modeling (6) indicates that the highest tensile stresses are in the region of the commissures

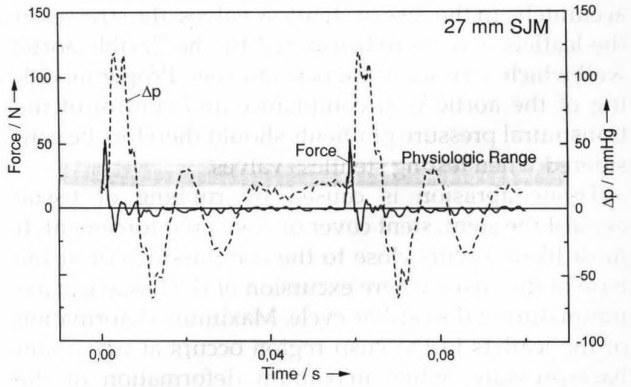

Figure 10: Integral loading force and pressure difference across valve for a 27 mm SJM valve at a test rate of 1000 per min, but at a preselected pressure difference of 120 mmHg (FDA-condition) across the valve.

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