A STUDY OF DISCHARGE COEFFICIENT IN BILEAFLET VALVES

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***Abstract* - The measurement of a cardiac valve's area is a common procedure, usually performed with noninvasive, Doppler-based techniques. Such measurements are not, however, without problems: a potential source of errors is the value of a valve's discharge coefficient.**

In-vitro pressure and flow measurements relative to the bileaflet valve of four brands were performed. A total of 12 valve samples was studied to cover the entire range of valve sizing. The data were used in the Gorlin formula for valve area measurements, and the dependence of the discharge coefficient on the internal diameter of the valve and the flow rate was accurately determined.

The reported results can be used in a great number of follow-up clinical assessments to improve the accuracy of valvular orifice measurements.

Key words - mechanical heart valves, hemodynamics, in vitro testing

I. Introduction

Since many years the quantification of the possible restriction of a heart valve's orifice area due, e.g., to thrombus formation and/or calcification [1, 2, 3], has become a common medical practice. The measurement of the valve orifice area is usually based on Doppler-derived velocity data, particularly on the Gorlin formula for the assessment of valve area [4]; with such techniques, the state of an implanted or natural valve can be monitored noninvasively.

We define, as usual, the discharge coefficient C_d of the valve as A_{eff}/A, A_{eff} being the cross-sectional area at the vena contracta (also termed effective area) and A the valve's cross-sectional orifice area (A). The Gorlin formula states that

$$A = \frac{Q}{51.6 C_d \sqrt{\Delta p}} \tag{1}$$

where Q is the flow rate in ml/s and Δp is the maximum pressure drop across the valve (measurable at the vena contracta) in mmHg.

To study the heart valve orifice area, the use of the Gorlin formula requires that C_d for each valve be known, in order to calculate the actual orifice area. This is a frequently overlooked problem, affecting the accuracy of orifice measurements. In-vivo C_d is often assumed to be equal to 1, especially in aortic position, as a simplifying approach. This is due also to the scarce information available on the discharge coefficient of PHVs; anyway, this measurement can be of interest, since letting C_d =1 means that the cross-sectional area at the vena contracta, A_{eff}, is being measured, instead of the true orifice area. The present study characterizes the principal parameters which determine the C_d value in bileaflet valve prostheses. The experimental data consisted in pressure recordings at several flow regimes in stationary conditions, for each valve. Bileaflet valves of four brands were tested, with three nominal size for each valve type; the ensemble of the data relative to a given valve type was then used to derive the functional relationship of the C_d with the Reynolds number and the dimensionless valve's internal diameter. The reported parameters characterizing this relationship can be used to improve the accuracy of valvular area measurements in implanted patients, by echographic equipments.

II. Materials and Methods

A. Experimental

Four bileaflet-type valves (Sorin Bicarbon, St. Jude HP, Edwards Tekna and CarboMedics, which will be referred to as SB, SJ, ET and CM, respectively), each tested in three sizes (19-, 23- and 27-mm TAD), for a total of 12 valves, have been studied in a steady-flow tester, at several flow rates, ranging between 5 and 35 l/min. The geometric data relative to each valve type and nominal size considered in the present study are reported in Table I: orifice inner diameter, mm and (in brackets) orifice area, cm^2. The data are those provided by the respective manufacturer.

A steady-flow apparatus was built, in which the flow was driven by a constant head tank. As reported in Fig. 1, one upstream and 16 downstream pressures were measured at equispaced locations (intervals of 17.5 mm, equal to D/2, where D is the diameter of the cylindrical inflow and outflow sections). Measured transvalvular pressures were interpolated in order to quantify the maximum pressure drop at the vena contracta.

TABLE I

	orifice inner diameter (mm) (orifice area (cm^2))			
TAD	SB	SJ HP	CM "R"	ET
19	15.2 (1.76)	16.7 (2.1)	14.7 (1.59)	14.1 (1.5)
23	19.24 (2.83)	20.4 (3.1)	18.5 (2.56)	17.5 (2.4)
27	23.30 (4.14)	24.1 (4.41)	22.5 (3.84)	21.5 (3.6)

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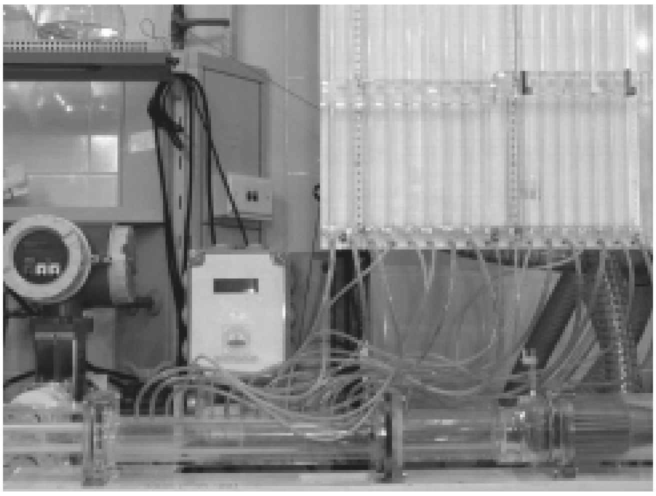

Fig.1 Experimental setup. The PHV seating is shown, together with the electromagnetic flowmeter (EMF), and the pressure measurement devices (P)

Flow rate was measured with an electromagnetic flowmeter (<0.1% accuracy) by Endress + Hauser (Atlanta, Georgia, U.S.A). A more detailed description of the apparatus can be found in [5]. The recorded data were used according to the procedure explained in the following section.

B. Characterization of the discharge coefficient

The Gorlin area (1) relies upon the precise determination of the valve's discharge coefficient C_d. It is a common practice to assume C_d =1 [6], especially for aortic valves, whereas a lower value (around 0.7) has been proposed for valves mounted in the mitralic position, on account of the larger upstream reservoir. Actually, when the flow must pass through a restriction, the contraction of the flow is more marked with increasing area of the upstream section; in the mitral position, then, a valve will have a smaller discharge coefficient (or A_{eff}/A ratio) than in the aortic position.

Using C_d =1, however, entails that the area being measured with the Gorlin formula is that of the vena contracta A_{eff}, not the actual valve area *A*.

Another technique often used to derive the dimensions of the valvular orifice from Doppler measurements is based on the continuity equation [6]

$$A_{CE,1} = \frac{Q_{peak}}{V_{peak}} \tag{3}$$

or

$$A_{CE,2} = \frac{\int Q(t)dt}{\int V(t)dt}, \tag{3}$$

where the velocity V_{peak} and $V(t)$ are the average velocities in the section, at peak systole or the time-dependent value (to be integrated on the systolic interval), respectively.

Actually, this definition of the valve area should not be termed "continuity equation", since it is based on the definition of the mean velocity in a section, while the continuity equation accounts for the conservation of mass in a given control volume.

If the peak values of the volumetric flow rate and the velocity (or, alternatively, the stroke volume and the time integral of the velocity) can be measured independently, this approach provides an estimate of the area of the section of the vena contracta. Therefore, the problem of calculating the real orifice area (taking account of the discharge coefficient) is not solved satisfactorily, in the same way as using the Gorlin equation with the hypothesis C_d =1.

As reported by several authors (e.g., [7]), the value of C_d can be substantially lower than unity (a value lower than 0.7 was reported for rigid, axisymmetric stenoses in [8]).

Furthermore, C_d can not be considered as constant, because the streamlines exiting the valve and slightly directed towards the orifice axis, redistribute themselves as the Reynolds number (Re) increases, with a consequent flow dependence of C_d.

Theoretically, it can be stated that C_d is dependent on the ratio between the area of the orifice and the area of the section of the upstream reservoir [9]. An accurate evaluation of these effects must be made, in order to derive realistic values for the discharge coefficient C_d. Therefore, we used the recorded pressure and flow rate data to calculate C_d for each valve type, choosing the ratio between inner orifice/inflow section diameter (d_i / D) and the Reynolds number (Re=VD/ν, V and ν being the mean velocity and the fluid kinematic viscosity, respectively) [9, 10] as principal parameters. According to the hypothesised functional relationship we assume the following power-law function:

$$C_d = C\left(\frac{d_i}{D}\right)^{\alpha}\left(\frac{VD}{\nu}\right)^{\xi} = C\left(\frac{d_i}{D}\right)^{\alpha}\left(\frac{4Q}{\pi D\nu}\right)^{\xi}. \tag{2}$$

Substituting (2) for C_d into the Gorlin formula (1), the least-squares fit of the experimental transvalvular pressure and flow rate data yielded the three constants C, α and ξ for each of the four valve types studied, thereby characterizing each valve by means of its discharge coefficient C_d, given by (2).

III. Results

A typical pressure-longitudinal distance profile is plotted in Fig. 2. The effect of pressure recovery is clearly visible, after the pressure minimum occurring slightly beyond a distance of one diameter downstream of the valve. The interpolated pressure minima were collected in an array and used, together with the flow rate data, to derive the parameters qualifying the C_d, according to (2).

TABLE II

	SB	SJ	CM	ET
C	0.6665	0.4914	1.3234	0.4148
ξ	0.0162	0.0664	-0.0500	0.0767
α	0.0189	0.4813	0.4045	0.4097

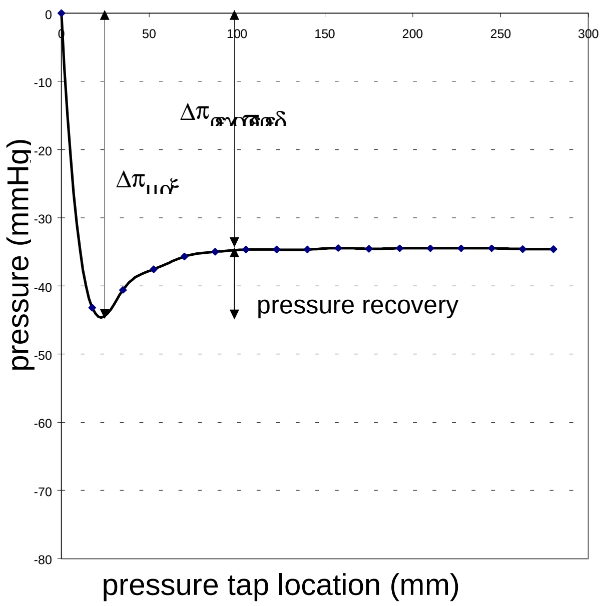

Fig.2. Longitudinal profile of the pressure drop for the 19-mm St. Jude HP tested in steady flow (30 l/min). The quantities Δp_{max} and $\Delta p_{recovered}$, as also the pressure recovery are shown

The variation of the C_d with flow rate is exemplified in Fig. 3 for the case with TAD=19 mm. The graph was plotted according to (2), with D equal to the diameter of the inflow section used in the measurements (D=35 mm). The flow rate is referred to the typical kinematic viscosity of blood (ν=3.7 cSt), taking account of the definition of the Reynolds number as $\mathrm{Re} = \dfrac{4Q}{\pi D \nu}$ (for the same Reynolds number, the flow rate can be scaled for viscosity as $Q_{eq} = Q_{meas} \dfrac{\nu_{blood}}{\nu_{meas}} = = 3.7 Q_{meas}$).

The SB valve shows the largest C_d of all the considered 19-mm-TAD bileaflet valves (Fig. 3).

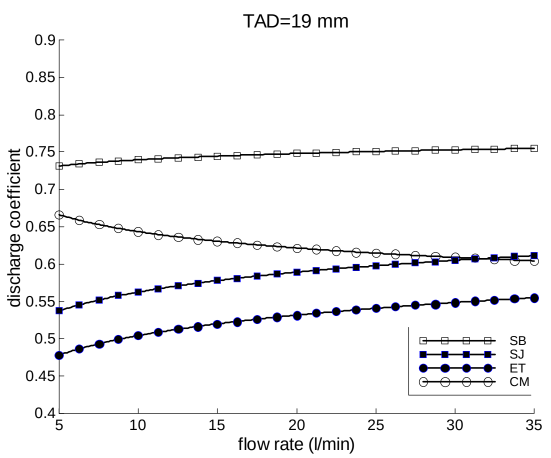

Fig. 3. Discharge coefficient profiles for the tested PHVs. The valve nominal size is 19 mm

The large SB's C_d is remarkable, since SB has a smaller orifice area (1.76 vs. 2.1 cm^2) than the SJ valve of the same nominal size, as evidenced in Tab. I. Also in Fig. 3 the similar flow dependence of SJ and ET as for the C_d can be appreciated, with SJ having an almost constant advantage over ET in the considered range of flow rates. Similar shapes of the C_d-flow rate relationship were found for the other valve sizes tested.

The CM valve is the only PHV to have a negative value of ξ (Table II), with the consequence that its C_d (hence, A_{eff}) decreases with flow rate (Fig. 3), as opposed to the other valve types (and also to simple stenotic orifices [8]).

IV. Discussion

In previous works, the comparison of measurements of PHVs in pulsatile and steady flow showed similar velocity profiles [12, 13], at an equivalent Reynolds number. In view of the similar behaviour of PHVs in steady flow and in peak phase of pulsatile flow, the values of the C_d we report are applicable to the cardiac phase characterized by the same flow rate. Orifice areas are determined over the systole or the central part of the latter, corresponding to the maximum instantaneous flow rate, when the valve is completely opened and considering a sinusoidal shape of the systolic aortic flow rate (assuming a systole/cycle ratio of 35%),

$$\overline{Q} = 0.35 \frac{2}{\pi} Q_{pk} = 0.2228 Q_{pk} .$$

For a 6-l/min cardiac output CO (or $\overline{Q}$), the peak flow rate can be calculated as $Q_{pk} = \overline{Q} / 0.2228$ =26.928 l/min, or 448.8 ml/s. Hence, the region of higher clinical interest for the flow regime is that in the 20-30 l/min range.

In the past, it has been speculated that the dependence of the discharge coefficient on flow rate (Q) could be due to an actual increase of the prosthesis' diameter with the flow rate, but this effect should appear only in biological valves, and much less (if any) in mechanical valves. Actually, it has been shown [8, 16] that even for rigid, axisymmetric stenoses the effective valve area at the vena contracta A_{eff} (hence, also $C_d = A_{eff} / A$) can not be considered as constant with Q, in spite of the fixed orifice diameter. Therefore, there is clearly an important role of the hydrodynamics in determining the effective area of a heart valve; this entails that each valve type must be accurately characterized, to account for the different hemodynamical performances arising from the subtle differences in design.

From Table II, it can be seen that CM is the only valve to provide a negative exponent ξ in (2). Also in [10] a positive correlation between systolic flow rate and C_d was found in mechanical PHVs. CM's behaviour is unexpected since, as the fluid is given more kinetic energy, usually it is more resistant to the driving effect of the restriction represented by the valve's ring and occluders, which moves the fluid toward the jet axis as it travels along a constriction, hence A_{eff} should increase [11]. In the case of CM, instead, the expected

increase of the area of the vena contracta does not occur; on the contrary, the restriction is enhanced by increasing Reynolds numbers. This could be related to turbulence production and/or fluid dynamical instability. Actually, previous studies of highly accurate velocity measurements, performed with laser Doppler anemomentry, have highlighted the loss of the central jet in the mean velocity profile downstream of the 19-mm-TAD CM, in both mitral [14] and aortic positions [15], probably due to leaflet fluttering. This interaction between flow and valve structure can explain why the streamlines (as far as this concept can be applied in a situation affected by turbulence inception) exiting the CM valve do not expand for increasing flow rates.

A great number of valvular orifice measurements has been reported, both in vitro and in vivo. In [6], for instance, the value of 0.94 cm^2 is reported for the Gorlin area of the 19-mm St Jude Std. valve, derived from catheter pressure measurements; the Doppler-derived Gorlin area, instead, was 0.71 cm^2. The catheter areas are always larger than the Doppler areas, since the latter can be affected by spatial averaging of the velocities in the insonated fluid volume; this effect tends to lessen the velocity peaks, biasing the area estimates towards larger values.

These values refer to the value 51.6 used in the denominator of (1). Since Baumgartner et al. [6] used C_d =1, these values can be regarded as estimates of the area of the section of the vena contracta.

Using the flow rates used in [6] to calculate the orifice area we computed the average of C_d over these values ($\overline{C_d}$ = 0.5790, from (2)); the same parameters of the SJ HP were used, since the similarity in design to the Standard type, tested in [6]. With this C_d value, the geometric orifice area could have been calculated, together with the mentioned result of the catheter Gorlin area relative to C_d =1, as A=0.94/0.5790=1.624 cm^2, vs. an internal orifice area, calculated with the ID (=14.7 mm) of the 19-mm St. Jude Std., of $(\pi/4)1.47^2$ = 1.697 cm^2. The relative difference between the two values is only 4.3%. This result indirectly confirms the validity of our findings.

V. Conclusions

The results of this study (see the functional parameters listed in Table II) show that each valve type has its own C_d value, at each flow rate, and its own relationship with geometrical parameters (internal diameter and inflow section diameter).

The proposed formula for the C_d enables one to take into account the dependence on flow rate of the effective area (i.e., the area of the vena contracta), which is commonly measured with ecographic techniques. The geometric area of the valve can thus be reliably measured, since the parameters affecting Doppler valve area estimations are correctly taken account of.

Valve manufacturers usually provide surgeons with the Gorlin valve area, as the effective area of their product, in view of making clinical orifice assessments easier. As shown in the present study, however, this parameter is not univocally determined, on account of the variation of the effective area A_{eff} (directly proportional to C_d) with geometric and fluidodynamical (Reynolds number) conditions. Using the formula (2) for the C_d, with the appropriate set of parameters, the actual orifice area (internal orifice area) can be measured with accuracy, regardless of the particular conditions in which the assessment is performed.

In conclusion, the knowledge of the above-mentioned functional relationships qualifies the performance of each PHV design, suggesting the possibility to use a more physically realistic discharge coefficient in the echographic protocols aimed at the assessment of valvular area.

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