Editorial Comment

Further Reconciliation Between Pathoanatomy and Pathophysiology of Stenotic Cardiac Valves*

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The applicability of basic hydraulic equations for calculating the orifice size of cardiac valves has been intensely studied and its clinical accuracy both affirmed and denied. One vexing issue has been the quantification of the empirical constant (C), which relates pressure and flow data to the anatomic valve orifice size. The value of C is composed of several elements: the hemodynamic elements are Cc (coefficient of orifice contraction) and Cv (coefficient of flow velocity). The former assumes that “there will be a contraction of the issuing stream such that the area of the stream [is] . . . less than that of the orifice itself” (1), whereas the significance of Cv is that “only . . . a fraction of pressure is converted to velocity. The rest is dissipated as frictional losses, turbulence, etc.” (1). Additional components of C are the conversion of fluid measurements from mmHg to cm H2O, and the derivation in 1951‡ of true diastolic filling period.

The present study. The precise impact of changes in orifice size and shape on the several constants has never been well studied. In this issue of the Journal, Flachskaempf et al. (2) have done just such a study with an elegant model system. They have reconfirmed that the larger the ratio of orifice perimeter to orifice area, the smaller the orifice constant, as exemplified with smaller areas and eccentric shapes, whereas a nozzle configuration of the inlet enhances the efficiency of the outflow. Furthermore, the authors (2) demonstrate that flow rate, except at very low levels, has no effect on orifice discharge efficiency. They also counter the experiments of Cannon et al. (3), which suggested that flow varies directly with the pressure difference across the orifice (or . . . change in discharge coefficient). Flachskaempf et al. (2) conclude that there is an 8.9% increase in the discharge coefficient with an eightfold increase in orifice area (from 0.3 to 2.5 cm²), a 5.9% decrease in the coefficient with a fivefold increase in eccentricity from a circle to an ellipse, and an 8.8% increase in the coefficient with a change from an abrupt to a nozzle-like orifice. The mathematical presentation is precise and long overdue. The major issue now becomes whether these refinements are of practical importance in the clinical laboratory.

Clinical applications. Let us examine specific examples: When the mitral valve is stenotic, a nozzle effect is likely as a result of the gradual narrowing of the fused leaflets from annulus to orifice. The shape of the orifice may vary from elliptic to circular, with respective discharge coefficients of 0.73 and 0.75. If the area of a circular orifice changes from 1 to 2 cm² with an intervention such as balloon dilation valvuloplasty, then the discharge coefficient changes from 0.75 to 0.79. The discharge coefficient of a 3:1 ellipse enlarged from an area of 1 to 2 cm² would change from 0.73 to 0.75. Because the orifice shape generally remains the same before and after valvuloplasty (4), only a correction for increase in orifice area is required.

The stenotic aortic valve, although having a nozzle-like inlet from the outflow tract to the valve itself, presents a fixed resistance perpendicular to the outflow lines. Furthermore, the orifice tends to be highly eccentric and often stellate with an excessive perimeter/area ratio. In this respect, the study of Flachskaempf et al. (2) would predict large pressure losses with a low discharge coefficient.

Patient size may also play a role, because smaller valve sizes are consistent with the lower cardiac output required. A different discharge coefficient may be applicable in children compared with adults as a result of the smaller actual size of the valve obstruction.

When hydraulic data are used to calculate valve area, the problem is complicated by the loss of energy that is expended in spreading and holding open the stiff, resistant leaflets and as blood flows past irregular surfaces. These factors augment the pressure difference across the valve. Nevertheless, the pressure/flow ratio accounts for the interaction of all factors: pressure losses resulting from all impedances, orifice size, shape and discharge contraction changes. Two-dimensional echocardiographic imaging of the mitral valve depicts the anatomy of the valve, whereas the hemodynamic calculations indicate its functional status. However, these two calculations may differ, not only at rest but also under diverse hemodynamic conditions. In practical terms, an intervention would do little for the patient if it...
improved valve area in the two-dimensional image without improving pressure/flow relations across the valve.

Finally, other inaccuracies and imponderables confound the hydraulic calculations. With current techniques, measurement of cardiac output has an error of ±5% at best, whereas pressure readings can err depending on the methods used and the site of recording (5,6). For example, some inaccuracy results from the use of pulmonary wedge pressure rather than left atrial pressure; and the pressure gradient, especially at low values or when not persistent throughout the period of flow, is best determined as the time integral (7). The imponderables arise from the fact that there may be other sources of impedance in the valve region, such as calcific intrusion, distortion or roughening of valve leaflets or additional subvalvular mitral or aortic stenosis.

The work of Flachskampf et al. (2) provides a quadratic equation by which the "true" valve area can be derived from the relative severity of stenosis and from the shape of the orifice once these values are known. This can also be derived by estimation. For example, if the area of an orifice is calculated as ≤1 cm², then it is probably about 5% larger. Similarly, an eccentric opening may be 5% larger than calculated. This hypothesis could be tested by comparing a carefully measured series of echocardiographic valve images with valve areas simultaneously calculated by hemodynamic methods.

Conclusions. The logical extension of the work of Flachskampf et al. (2) is that the cardiologist should estimate valve obstruction by both anatomic and physiologic methods. The two data sets may not always match, because they should not necessarily match! The use of correlation coefficients that force the data to fit or not fit may no longer be the proper exercise. Unusual dissipation of pressure that yields a discrepancy in derived valve area (8) may be telling us more about the valve than we knew before. The discrepancy may reveal that there are additional sources of energy loss, such as valve leaflet resistance to bending with onset of flow, and irregularities of the valvular or perivalvular surface causing frictional losses and possibly eddies and whorls. The hydraulically derived area may characterize the total impedance offered by the flow path. We need not be frustrated when valve areas derived by two different methods are not the same. Rather, we should, like the princes of Serendip, seize upon the difference for what it reveals about the anatomy of the valve orifice as opposed to the physiology of the total obstructing valve tract under active pressure and flow conditions.

References