Dole et al. (1984) have suggested that the coronary zero flow pressure results from critical closure of microvessels, which they model as pressure-dependent nonlinear resistors. This view differs from the vascular waterfall interpretation of coronary hemodynamics in which microvessels are modeled as collapsible tubes. It is important to determine which model is more descriptive, for an understanding of the back pressure and resistance of the coronary circulation.

In the case of a pressure-dependent nonlinear resistor, what is the relationship between cross-sectional area and pressure, i.e., its tube law? The solid line in Figure 1 is a plot of the inverse of the square root of the resistance function shown in Figure 10 of Dole et al. (1984) as a function of epicardial artery pressure. This is equivalent to a tube law, since the ordinate is proportional to cross-sectional area. However, the tube law at the site of critical closure will differ, because it will be displaced to the left by resistive losses. It may resemble the dashed line; if so, the marked decrease in cross-sectional area caused by a slight decrease in pressure is similar to what has been found for collapsible tubes.

If a pressure-dependent nonlinear resistor can act like a collapsible tube, why isn't coronary flow independent of venous pressure perturbations? There are two explanations as to why upstream and downstream hydraulic events are uncoupled when an intervening collapsible segment exists. First, the velocity of fluid flow through the collapsible segment exceeds the speed with which a disturbance can propagate in the upstream direction (Griffith, 1971). Second, the hydraulic resistance of the collapsible segment automatically adjusts to changes in downstream pressure so as to keep flow constant (Rubinow and Keller, 1972).

The first explanation is unlikely, because wave speed in the microcirculation almost certainly exceeds the velocity of fluid flow (Caro et al., 1978). The second explanation requires that the uncoupling be relative and dependent upon the magnitude of the ratio of the squared cross-sectional areas of the collapsible segment in its narrowed and open configurations. Only when there is little flow will downstream pressure not be a determinant of flow. Therefore, even if coronary microvessels can act as collapsible tubes, flow need not be independent of venous pressure perturbations.

Even allowing for vascular waterfalls in individual vessels, the bed as a whole may not show waterfall-like hydraulics. Depending on the magnitude of the pressures tending to obliterate or to maintain the lumen (the surrounding and intraluminal pressures, respectively), a collapsible tube can exist in several configurations: open, closed, or in the transitional state between these two configurations. A vascular waterfall is possible only in the transitional state. Given a population of collapsible vessels with different surrounding pressures, all three configurations may coexist. If the pressure range over which the transitional state exists is small, compared to the range of surrounding pressures, most vessels will be either open or closed.

Figure 2 shows an analysis of flow through multiple parallel tubes with collapsible segments that have different surrounding pressures. The frequency distribution as a function of surrounding pressure is approximately log normal. The thin lines are composite pressure-flow relations for all tubes with a specific surrounding pressure. The pressure range of transitional state is 5 mm Hg, and the surrounding pressure span is 25 mm Hg. The composite pressure-flow relation for all collapsible segments is shown as a broken line, while the upstream relation is the thick line to the right. The latter relation was obtained by assuming that the tubes upstream from the collapsible segments offer constant resistance to flow.

When upstream pressure is 100 mm Hg, the pressure at the level of the collapsible segments is 37 mm Hg (arrow). Flow occurs only in tubes with a surrounding pressure equal to or less than 37 mm Hg. Waterfall-like hydraulics are possible only for pressure-flow relation number eight. Thus, the back pressure for almost all flow is the outlet pressure.
FIGURE 2. A "thought experiment" dealing with flow through multiple parallel tubes containing collapsible segments with different surrounding pressures. Dots indicate the operating points on pressure-flow relations of perfused tubes when steady state conditions are: inflow pressure, 100 mm Hg, and flow: 100 ml/100 g per min. The collapsible segments described by pressure-flow relations numbered 1-7 are in the open configuration; the outflow pressure is the back pressure. A vascular waterfall exists only in number 8; the surrounding pressure is the back pressure. Surrounding pressures are also the back pressures in all tubes to the right of the arrow, but, since none of these tubes have flow, these back pressures are "hidden" and do not affect the upstream pressure-flow relation.

not a surrounding pressure. Flow is controlled by resistance calculated as total pressure drop divided by flow, and not as the reciprocal of the slope of the upstream pressure-flow relation.

Certain conclusions can be drawn from this "thought experiment."

1. Dole et al. (1984) are correct in asserting that the coronary back pressure is venous pressure, and that resistance should be calculated as the difference between arterial and venous pressure divided by flow.

2. The coronary zero flow pressure indicates only the lowest surrounding pressure and is the back pressure for only a small portion of the bed, and, then, only when flow has ceased or is about to cease.

3. Critical closure may occur in collapsible vessels without waterfall-like hydraulics being conspicuously apparent.

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References

Reply to the Preceding Letter
Recently, we suggested that diastolic coronary artery pressure-flow relationships and zero flow pressures obtained during steady state conditions could be explained by a model incorporating a pressure-dependent nonlinear resistance analogous to the phenomenon of critical closure (Dole et al., 1984). As Dr. Bellamy has indicated, this view differs from the vascular waterfall model of coronary hemodynamics which he originally proposed to explain diastolic coronary artery pressure-flow relationships (Bellamy, 1978). Whereas both models predict positive zero flow pressures ($P_{fo}$) greater than venous pressure ($P_v$), the hemodynamic factors regulating flow are quite different. In our model, the back pressure opposing flow is $P_v$, and flow stops at $P_{fo} > P_v$, because total coronary resistance approaches infinity as inflow pressure approaches $P_{fo}$. In the waterfall model, the back pressure opposing flow is $P_{fo}$, which remains constant despite changes in flow or $P_v$ for $P_v \leq P_{fo}$. The zero flow pressure is regulated by a collapsible vascular segment whose geometric configuration is passively determined by transmural pressure (Holt, 1958; Permutt and Riley, 1963; Conrad, 1969). Although the relationship between vascular cross-sectional area and pressure (tube law) predicted by our model appears to be similar to what has been described for collapsible tubes, an important difference is that our model does not contain a resistive segment which automatically adjusts to changes in flow or $P_v$ to keep pressure just upstream to the segment constant at $P_{fo}$.

In previous publications, Dr. Bellamy had assumed that the vascular waterfall phenomenon occurred over the entire physiological range of pressures (Bellamy, 1978, 1980; Bellamy et al., 1980). In
his letter, he has proposed that waterfall hemodynamics may be limited to a very narrow pressure range compared to the range of pressures surrounding the collapsible vessels. Since most vessels will be either open or closed, the back pressure for flow is \( P_v \) and not \( P_{f_0} \). In his analysis (Fig. 2), Dr. Bellamy assumes that the frequency distribution for collapse of individual segments is approximately log normal over a 25 mm Hg pressure range and that the total resistance upstream from the collapsible segments is constant. Since the back pressure opposing flow for all vessels is \( P_v \), when inflow pressure is greater than the maximum surrounding pressure (50 mm Hg in Fig. 2), it follows that the pressure-flow curve will be linear above 50 mm Hg, and that this segment, when extrapolated to the pressure axis, will intersect at \( P_v \) not at \( P_{f_0} \) as indicated in Figure 2. That is, the resultant sum of lines 1–8 (composite pressure-flow relationship) above 50 mm Hg is a line whose slope is the sum of individual slopes and whose intercept is \( P_v (P_f = 0 \text{ in Fig. 2}) \). As pressure is reduced below 50 mm Hg, the pressure-flow relationship will become steeper and could be nonlinear. Although a distinct inflection point may not be evident because of the frequency distribution of surrounding pressures, a concave bend in the curve toward the pressure axis intercept \( P_{f_0} \) must occur. The absence of such curvature in published steady state diastolic pressure-flow relationships argues against Bellamy's proposal unless one also assumes that changes in resistance occur even in the higher pressure range.

The effect of transmural pressure on coronary vascular resistance has recently been examined by Hanley et al. (1984) in the isolated empty, beating, maximally dilated dog heart. These investigators found an inverse relationship between inflow pressure and resistance. Since this relationship was uniform across the myocardial wall, sequential drop-out of vessels by transmural layers cannot explain the dependency of resistance on pressure under the experimental conditions. It was suggested that changes in vessel radius and/or vascular recruitment may explain the inverse relationship between pressure and resistance.

An important observation which is not readily explained by a passive collapsible tube model is the dependency of compliance-free estimates of \( P_{f_0} \) on vasomotor tone (Klocke et al., 1981; Dole and Bishop, 1982; Lee et al., 1984). Whereas the theory of critical closure predicts that \( P_{f_0} \) will depend on active vascular smooth muscle tension, it is not apparent how active vasomotor tone would affect the hemodynamic characteristics of a collapsible tube.

We appreciate Dr. Bellamy's comments and agree with his conclusion that the coronary back pressure is \( P_v \) and that the total resistance regulating flow under steady state conditions should be calculated as the difference between arterial and venous pressure divided by flow. This necessarily implies that, in the absence of autoregulation, resistance is not constant during changes in coronary pressure, but increases as the pressure is reduced, consistent with the data of Hanley et al. (1984). Interventions which alter \( P_v \) do so, not by changing the back pressure to flow, but by altering the pressure-resistance relationship. Dr. Bellamy's distributed model with waterfall hemodynamics confined to a narrow pressure range, while theoretically possible, does not explain the slope of the pressure-flow curve above the maximum surrounding pressure. Direct proof of waterfall hemodynamics or of critical closure at or near \( P_v \) will require a better understanding of pressure regulation at the microvascular level.

Letters to the Editor

References


INDEX TERMS: Coronary pressure-flow relationships • Vascular waterfall
Comments on "Interpretation and physiological significance of diastolic coronary artery pressure-flow relationships in the canine coronary bed".

R F Bellamy

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