Effects of stenosis on coronary flow

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Principles of basic fluid dynamics

According to accepted laws of fluid dynamics applied to laminar, steady flow in an in vitro system of rigid tubes, the energy or pressure loss due to constriction of a tube is caused primarily by (a) viscous friction between layers of fluid in the stenotic segment according to the Hagen-Poisuelle Law and (b) flow separation or vortex formation (eddying or swirling) at the downstream end of the stenosis. The pressure loss, ΔP, is related to the flow velocity, V, through a stenotic tube according to a general equation, which may be written in simplified form as $\Delta P = FV + SV^2$. F is the coefficient of pressure loss due to viscous friction and is dependent on relative percent narrowing, absolute diameter, and length of the stenosis. S is the coefficient of pressure loss due to flow separation and is dependent on relative percent stenosis and exit or divergence angle of the stenosis. V is the first power of instantaneous, mean cross-sectional flow velocity, and V^2 is the second power of velocity. An additional term may be added in order to account for inertial losses associated with pulsatile flow. However, the inertial effects are small for stenoses above 50% diameter narrowing and can be omitted as applied to the coronary circulation. The fluid dynamic equations can be written as follows in terms

of coronary flow velocity or volume flow:

$$\Delta P = \frac{8\pi\mu L}{A_s}\frac{A_n}{A_s}V + \frac{\rho k}{2}\left(\frac{A_n}{A_s} - 1\right)^2 V^2$$

Velocity equation

$$\Delta P = \frac{8\pi\mu L}{A_s}\frac{1}{A_s}Q + \frac{\rho k}{2}\left(\frac{1}{A_s} - \frac{1}{A_n}\right)^2 Q^2$$

Flow equation

where ΔP is pressure loss across the stenosis, μ is absolute blood viscosity, L is stenosis length, r_s is absolute radius of the stenotic segment, A_n is the cross sectional area of the normal artery, A_s is the cross sectional area of the stenotic segment, V is flow velocity, ρ is blood density, k is a constant related to the exit or divergence angle, Q is volume flow. Blood viscosity and density are constant, and the relation between pressure loss and coronary flow or flow velocity is therefore a function of stenosis geometry. In the flow equation, $1/A_n$ is much smaller than $1/A_s$. Therefore, A_n or the normal size of the artery for modest narrowing has little influence on severity of stenosis in the flow equation. For that reason, changes in the normal artery diameter due to vasomotion can be ignored. The effects of stenosis in the flow equation are dependent primarily on the absolute diameter of the stenotic segment, not on relative percent stenosis or diameter of the normal adjacent artery, although the latter has some influence. These equations have been demonstrated to apply to a semi-in vitro model in which a machined plastic "stenosis" plug was inserted into the femoral artery of anesthetized dogs. The predicted pressure loss is accurate to within ±16% of true measured pressure loss. Severity of stenoses in the velocity equation is highly dependent on relative stenosis as well as on absolute stenosis diameter. Severity of stenosis with the velocity equations would therefore be sensitive to vasomotion of the normal part of the artery. With quantitative coronary arteriographic analysis, the general form of these equations has been proved to apply to intact, awake, chronically instrumented animals, but their quantitative accuracy for predicting in vivo pressure gradients has not been demonstrated yet. In addition, experimental data in models suggest that the details of stenosis geometry such as eccentricity, streamlining, or exit angle divergence and entrance shape are not of great importance. In simple terms, both equations show that the pressure gradient across a stenosis increases sharply and in a progressive curvilinear fashion with increases in coronary flow. Therefore, the effects of a stenosis will be least at resting coronary flow and greatest at high coronary flow.

Effects of stenosis on systolic and diastolic coronary flow

During progressive coronary constriction under resting conditions, diastolic flow decreases and systolic flow increases until the characteristic phasic pattern of coronary flow damps out with little fall in mean flow. Thus, paradoxically, systolic coronary flow increases with progressive stenosis. Further constriction causes a fall in mean flow. When mean flow is reduced to half or less of control values, the phasic pattern of coronary flow resembles that of aortic pressure with systolic flow being higher than diastolic flow. For a moderately severe stenosis (up to 80% diameter narrowing 3 mm or less long) at rest, the pressure gradient is small and flow shows the

characteristic phasic pattern of coronary flow. During coronary vasodilation, the pressure gradient becomes more severe as flow increases, the phasic pattern of coronary flow in the presence of a fixed, constant stenosis becomes damped during peak flow after a vasodilatory stimulus, and the phasic pattern returns as flow returns to baseline. Systolic flow increases proportionately more than diastolic flow during vasodilation until there is little phasic variation. Therefore, stenoses have different effects on systolic flow than on diastolic flow in the coronary circulation.

Changes in stenosis geometry after coronary vasodilators

In the above equation, $\Delta P = FV + SV^2$, the linear first term due to viscous friction accounts for 65% and the nonlinear term due to flow separation accounts for 35% of the total pressure loss at resting coronary flow in experimental animals. At peak coronary flow after coronary vasodilation, the pressure loss due to viscous friction accounts for 33% and pressure loss due to flow separation accounts for 67% of the total pressure loss. The pressure gradient-velocity relationship at high flows is characterized by the same general equation but with proportionately larger values of the coefficient S and therefore greater pressure loss associated with flow separation than predicted by the resting gradient-velocity relationship. The pressure loss predicted for high coronary flow velocities on the basis of the gradient-velocity equation at rest is only 64% of the actual experimentally observed pressure gradient at peak coronary flow. The augmented separation loss following coronary vasodilation may be due to dilation of the epicardial artery adjacent to the fixed stenotic segment, which causes more severe relative percent narrowing and a lower flow velocity in the normal segment of artery. The average increase in diameter of the normal coronary artery adjacent to the stenosis is 30% to 50%. If absolute volume flow measurements were made, then there would be little effect of adjacent epicardial vasodilation on flow since relative percent stenosis is of little importance in the flow equation, but absolute stenosis diameter is the dominant factor. In the presence of severe stenoses that reduce resting coronary blood flow, vasodilators result in a paradoxical further fall in flow associated with a marked increase in stenosis resistance. This paradoxical fall in flow and increase in stenosis resistance has been ascribed to a fall in intraluminal pressure in the stenotic segment sufficient to collapse it, particularly in animal models in which a normal flexible vessel wall is externally compressed by a cuff constrictor in order to produce a stenosis. In human coronary arteries in which there is no external constriction, quantitative coronary arteriographic analysis shows that normal and stenotic segments of diseased arteries dilate after nitroglycerin, with stenosis cross-sectional area increasing 17% to 28% and with calculated stenosis resistance decreasing 20% to 26%. Thus, the overall functional geometry of coronary stenoses are dynamic, depending on different physiologic states and on how stenoses are produced.

Clinical and research applications

Long lesions. In experimental animals, coronary stenoses of 60% diameter narrowing 1 mm long have no effect, 10 mm long reduce coronary flow reserve significantly, and 15 to 20 mm long reduce resting blood flow. At low or resting flow the pressure gradient will increase linearly with stenosis length because most of the pressure loss is due to

the linear viscous term of the hydraulic equation, which is directly related to length. For example, doubling stenosis length will double the pressure gradient at low flows. However, at high flows, the pressure gradient does not increase in proportion to length because most of the pressure loss is due to localized turbulence at the distal end of the stenosis, a loss that is independent of length. In this instance, doubling the length may increase the gradient by only 10% to 20%.

Lesions in series. Pressure gradients and resistances of stenoses in series are directly additive, but the effects of constrictions in series on flow are complex. In some circumstances, this concept is accurate. For example, stenosis of 80% of arterial diameter does not reduce resting coronary flow; if in series with 95% stenosis, a degree of constriction that does affect resting flow, the more severe lesion determines entirely the effects on resting flow. Similarly, stenosis of 30% of arterial diameter does not reduce hyperemic response or coronary flow reserve; if in series with 60% stenosis, a degree of narrowing that does reduce hyperemic response, the more severe lesion also determines entirely the effects on hyperemic flow. However, if both lesions are separately severe enough to reduce flow, their effects in series are additive and the more severe lesion does not determine all the effects on flow. The hyperemic response of two 75% stenoses in series is less than that of one stenosis alone. Similarly, two 95% stenoses in series reduce resting flow more than one alone. Two 50% lesions in series will reduce coronary flow reserve even though neither of them alone will do so. Thus, series lesions cumulatively affect coronary flow reserve, resting flow, total stenosis resistance, and pressure gradient.

Quantitative coronary arteriography is unlikely to be useful in routine clinical practice because (a) coronary artery disease is often diffuse or a mixture of diffuse and discrete disease and the application of current analysis techniques have not been shown to apply to this situation, (b) the x-ray films must be of such high quality and the tracings of stenosis borders done with such precision that routine analysis of most clinical arteriograms will be difficult, (c) a computer, software, and a basic understanding of hydraulics are necessary to obtain and interpret quantitative coronary arteriography. However, for clinical research and animal experimentation, quantitative coronary arteriography is an essential and powerful tool. Due to its well-known variability, an "eyeball" estimate of stenosis severity is no longer an acceptable method for obtaining high quality scientific data.

Should percent stenosis or absolute stenosis dimensions be used to assess stenosis?

This question is really the same as asking if the velocity equation or the flow equation should be used? The advantages of using the flow equation are that (a) the equation is simpler, (b) it deals with volume flow, which physicians intuitively understand better than velocity, (c) a single measure of absolute stenosis diameter and length define the stenosis relatively independent of how big the normal artery is on either side of the stenosis; thus, there would be little change in effective stenosis severity with vasodilation of the normal segment of artery as long as the stenotic segment remained fixed. The disadvantage of the flow equation is that it does not provide a quantitative assessment of stenoses independent of knowing blood flow in the stenotic artery, a limitation that in some

comparative situations makes the flow equation appear erroneous from an intuitive, common sense point of view. For example, in absolute terms, a 2-mm diameter stenosis in a 4-mm diameter coronary artery is a 50% stenosis, which we know to be of mild physiologic significance. However, a 2-mm stenosis in the aorta having a normal diameter of 20 mm causes a 90% stenosis, which we know is a severe physiologic lesion. However, the geometric severity of the 2-mm stenosis, according to the flow equation, would be comparable in the two instances. We could interpret the significance of that geometric severity only if we knew it were a coronary artery or the aorta or, in other words, only if we knew how much flow should normally be going through the artery. Thus, use of the flow equation is of little value per se unless we also know what the volume of flow should be in the artery. The velocity equation does not have this problem because it includes a measure of the normal arterial size and because in the arteries of the body, mean cross-sectional flow velocities tend to be fairly uniform within a relatively narrow range regardless of the size of the artery (between 20 to 40 cm/sec at rest and four times that at peak; by comparison, resting volume flow ranges from very small values to 6000 cc/min in the aorta). Because of this fact, stenosis geometry defined by the velocity equation provides an assessment of severity without having to know which specific artery is involved or how much flow goes through it. This logic also explains why percent stenosis has served practically for so many years as an effective, intuitive measure of severity, even though the precise reason is not widely recognized. The disadvantages of the velocity equation are that it is more complicated, flow velocity is less intuitively understood, and stenosis severity will change dynamically with adjacent artery vasomotion.

Thus, to ask the question what percent stenosis affects flow is to ask an illogical question that cannot be rigorously answered. The proper questions would be (a) what is the absolute stenosis diameter that affects the normal volume flow in a specific artery, or (b) what is the percent narrowing and absolute stenosis in any artery that reduces its flow velocity?

In the author's opinion, the velocity equation is more general and broad based and therefore more useful for scientific purposes. It relates more closely to the use of percent stenosis than does the flow equation. However, the velocity equation also requires that we become accustomed to thinking in terms of flow velocity and dynamic changes in stenosis geometry.