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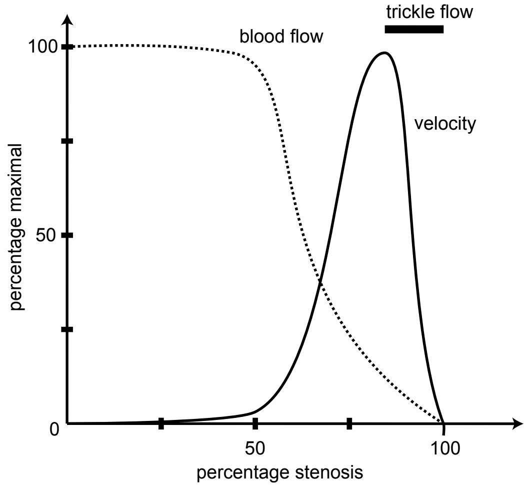

Figure 3. Flow and velocity changes expressed as percent maximal in response to percentage stenosis.

1.3.4 Disturbed flow

Laminar patterns of flow may be disturbed in the absence of disease in certain circumstances. As mentioned above, there is normally a 'boundary layer' of blood at the vessel wall which is in a 'no slip state'. At bifurcationsbranches or curves in the vascular tree, bizarre eccentric jets are formed which cause disturbed flow.

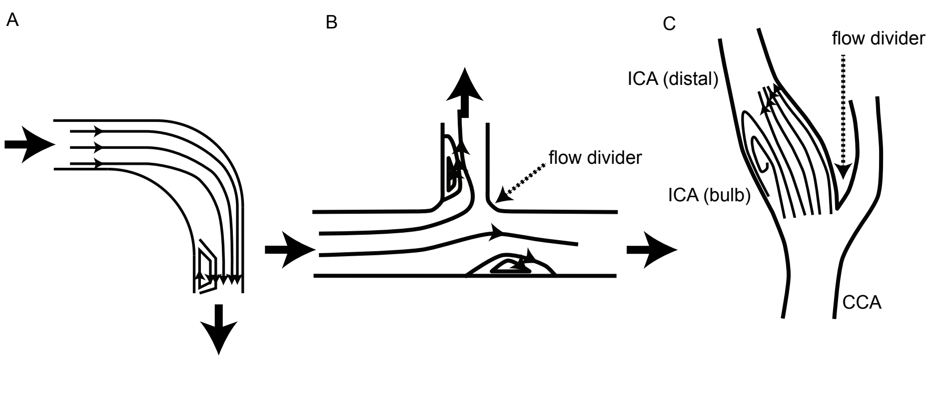

Figure 4. Schematics demonstrating patterns of flow A) around a bend B) at a branch vessel exiting at 90 δεγρεεσ, and C) in the carotid bulb. Note that the flow is fastest closest to where there is a flow divider. Boundary layer separation situations are seen where flow is stagnant.

Such effects are well recognised in the internal carotid bulb where an area of boundary layer separation occurs. Similarly at the inner corner of a branch artery which divides at 90 degrees from the main artery, the inside corner of a distal bypass graft anastomosis and the inside curvature of a curved vessel (see Figure 4)

No clear differentiation between such flow states and flow disturbance produced by local lumen irregularities is possible, other than by direct visualisation of the plaque.

1.3.5 Pulsatile pressure and pulsatile flow

The above discussion accurately describes constant flow in non-distensible vessels. The actual situation *in vivo* approximates to this when considering averaged pressures and velocities.

Clearly blood pressure generated by the heart is pulsatile. The steep systolic rise in pressure in arteries is due to blood being ejected from the heart faster than it can flow away. Much of this pressure is stored temporarily in the elasticity of the arteries. As systole wanes, outflow outstrips inflow and pressure decreases.

As the vasculature is not rigid, pressure increases take some time to translate through the vasculature. The velocity of this pressure or pulse wave ranges from about 5 to 15 m/second where as blood flow in the aorta is only about 0.2 m/second. This discrepancy between the pulse wave velocity and the blood flow velocity can be explained by understanding that blood is incompressible. Hence forwards motion is achieved by shunting the blood ahead and also by distending the vessel ahead.

The relationship between pressure gradients and flow in arteries was investigated by observing injected bubbles into the femoral arteries of dogs whilst measuring pressure from transducers inserted at two points in the vessel (McDonald, 1955).

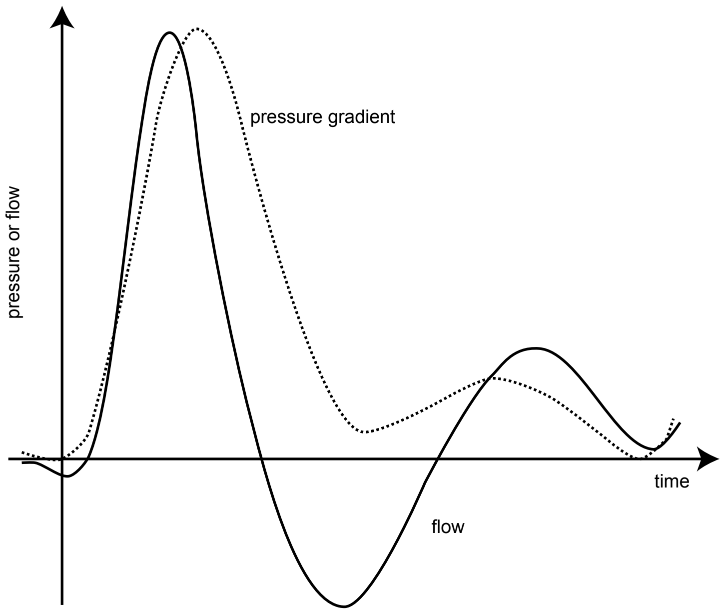

Figure 5. Relationship between flow and pressure in the femoral artery. Redrawn from McDonald, 1955).

McDonald observed that flow is consistently comprised of: fast forward flow during systole followed immediately by a back-flow phase, which in turn is followed by a forward flow in mid diastole, which is terminated by a brief back-flow phase. McDonald also observed that flow in the femoral artery oscillates in the same way as the pressure gradient but with a phase lag which varies throughout the cycle (Figure 5).

2) Absorption: acoustic energy is converted to heat energy. Under normal circumstances with diagnostic ultrasound, the amount of heat generated should be too small to generate a temperature rise.

Other causes of attenuation are divergence of the ultrasound beam and possibly refraction.

2.1.4 Specular and diffuse reflectors, Rayleigh scatterers

A specular reflector is a large smooth interface with dimensions much larger than the wavelength of the ultrasound. These sorts of reflectors produce strong ultrasound echoes in a single direction. As long as the reflected wave returns to the transducer, large amplitude echoes are produced.

Often surfaces in the body are rough. Such interfaces are known as diffuse reflectors. Echoes from diffuse reflectors are sent off in many directions due to the roughness of the interface. These echoes are hence easier to detect but may be of lower intensity.

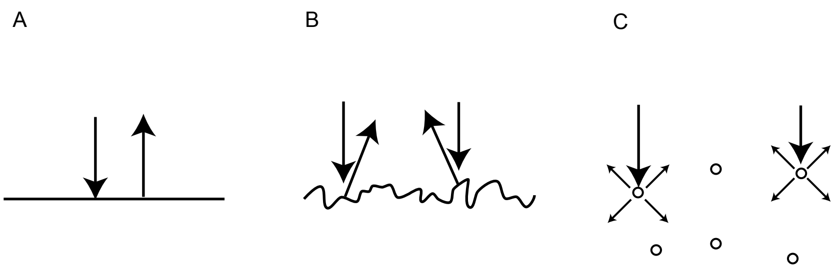

Figure 1. Illustration of A) Specular reflection from a smooth surface; B) Diffuse reflection from a rough surface and C) Scattering from Rayleigh scatterers.

For objects which are similar in size to the wavelength of the ultrasound, scattering of the incident ultrasound wave occurs. Hence such objects are termed scatterers.

For very small scatterers, the degree of scattering increases with increasing frequency of the incident ultrasound and with increasing size of the scatterer – i.e., reflection from scatterers is frequency dependant as opposed to reflection from specular reflectors.

Scatterers which are much smaller than the wavelength of the ultrasound (e.g., red blood cells) are termed Rayleigh scatterers. For these objects, scattering increases in proportion to the fourth power of the wavelength . Hence, if the frequency of the ultrasound is doubled the scattered power will increase sixteen times.

Note that the penetration (the maximum distance from which it is possible to detect scattered echoes) is greater at lower frequencies. This is so because attenuation, the reduction in amplitude with increasing distance travelled, is dependant on reflection and scatter at interfaces and also absorption. In most cases, attenuation is almost proportionate to frequency. Hence the lower frequency probe will result in less wave attenuation and hence greater penetration to reach small scatterers at greater depth.

2.1.5 Acoustic Impedance

The degree of reflection at any interface between two media is related to the characteristic properties of the two media. This characteristic property is called the acoustic impedances.

The specific tissue acoustic impedance (Z) is the relationship between the excess pressure (P) and the velocity (v) of particles in the medium:

$$Z = \frac{P}{v}$$

The impedance can be calculated from the product of the density of the material and the speed of sound in it (c):

$$Z = \rho c$$

Using $c = \frac{1}{\sqrt{\kappa\rho}}$ m

$$Z = \frac{\rho}{\sqrt{\kappa}}$$

It can be seen that the relationship between excess pressure and particle velocity is determined by the density and compressibility of the material.

2.1.6 Attenuation coefficient and Decibel

The attenuation coefficient is the degree of sound beam attenuation in a tissue, usually given in decibels per cm (dB/ cm).

In other words it quantifies the difference in signal amplitude received by a transducer between two similar reflectors one cm apart. The attenuation coefficient is roughly proportional to the frequency of the ultrasound wave.

The decibel scale is a way of expressing ratios of intensities or powers of signals.

- Intensity ratio (dB) = $10 \log_{10}\left(\frac{I}{I_0}\right)$ dB
- Power ratio (dB) = $10\log_{10}\left(\frac{P}{P_0}\right)$dB

Since the power is proportional to (voltage)^2:

- Signal ratio (dB) $= 10\log_{10}\left(\frac{V}{V_0}\right)^2$ dB

$$= 20 \log_{10}\left(\frac{V}{V_0}\right) \text{dB}$$

Where P_0 and V_0 are initial power and voltage and P and V are power and voltage at a second point.

Since it is a logarithmic scale, a wide range of signals are compressed into a relatively small scale. Also, due to the logarithmic nature, products of ratios in dB encountered when dealing with attenuation in different layers of tissue or successive stages of signal processing, may be added to yield total change in signal.

Example 1. Calculate the total attenuation in dB of an ultrasound beam of 3.5 MHz which is received from a target at a depth of 12cm in tissue with an attenuation coefficient of 0.8 dB/cm/MHz.

Total path length = 24 cm

Attenuation = 3.5 X 24 X 0.8 = 67.2 dB

Example 2. What is the difference in decibels between an intensity of 1 mWcm^{-2} and one of 2 mWcm^{-2}?

Relative intensity level (dB) = 10 log (I_2/ I_1) where I refers to intensity

Relative intensity = 10 log 2

= 3 dB

6.7 Algorithm for NASCET grading of ICA stenosis

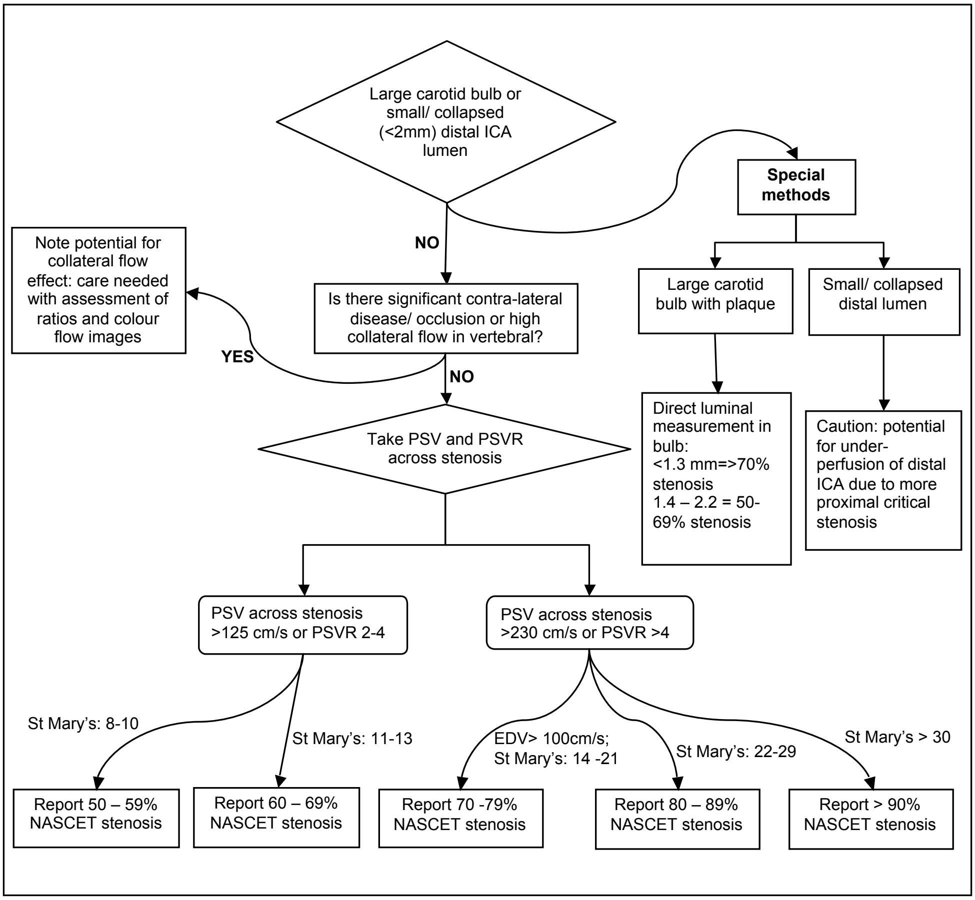

Figure 10. Decision making algorithm for diagnosis of carotid stenosis