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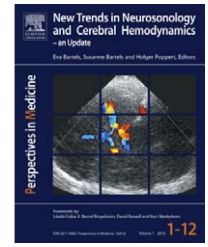

How local hemodynamics at the carotid bifurcation influence the development of carotid plaques

Dieter Liepsch[a,*], Andrea Balasso[b], Hermann Berger[b], Hans-Henning Eckstein[b]

[a] Insitut f. Biotechnik, e.V. Lothstr. 34, 80335, University of Applied Sciences, Munich, Germany
[b] Interdisciplinary Research Laboratory of the Klinikum rechts der Isar der TU München,Germany

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Patch plastic

Summary A short introduction is given of how fluid dynamics forces and velocity distribution influence the development of plaque in the carotid bifurcation. The flow parameters are discussed. Flow visualization techniques and also laser-Doppler-anemometer measurements demonstrate the importance of the flow. This will be shown in true-to-scale, physiological accurate models of the carotid arteries. These models have the same compliance as the real blood vessel. Some applications are shown e.g. patches, stents and filters. The most important factors are the flow rate ratio and geometry, unsteady pulsatile flow, wall elasticity and non-Newtonian flow behavior of blood.

Introduction

Cell/cell and cell/vessel wall interactions have been the subject of investigation and discussion for more than 40 years. It has been shown that low and high shear regions caused by flow separation regions and oscillatory flow are primarily responsible for chemical reactions which contribute to the formation of arterial plaques.

Most previous shear stress studies have only measured the axial velocity component at a few local points. They calculated the shear stresses with the velocity gradients using a constant viscosity. Accurate three-dimensional or, at least, two-dimensional velocity measurements are necessary to calculate the shear stresses. This is because, at bends and bifurcations, the secondary flow cannot be neglected. Numerical studies very often neglect the real, local viscosity of blood and the compliance of the vessel wall which shows a hysteresis. It is also very important that the non-Newtonian flow behavior of blood be considered, especially in flow separation regions.

We have studied the flow behavior in more than 200 arterial models with a different geometry and different flow rate ratios. The principles of hemodynamics, such as the forces on fluid elements, are important. These forces are

- the volume forces (such as gravity force or centrifugal forces);
- the pressure forces (normal pressure multiplied with the area of the fluid element);
- friction forces (shear stresses multiplied with the surface of the cell fluid element).

Normally, in larger blood vessels, the blood cells are concentrated in the center whereas the plasma flows near the

* Corresponding author at: Fakultät 05, Lothstr. 34, 80335 Germany. Tel.: +49 89 1265 1522; fax: +49 81573160.
E-mail addresses: dliepsch@t-online.de, liepsch@hm.edu (D. Liepsch).

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wall. The blood cells are deformed in capillaries where physical/chemical reactions take place. However blood cells are also occasionally transported into these recirculation zones in larger blood vessels, at bends and bifurcations. The cells remain in the recirculation zones over several pulse cycles and are subjected to both high and low shear stresses.

Many papers use the term 'turbulent flow', however a true turbulent flow is found only in the ascending aorta and this is not fully developed because of the entrance length. Everywhere else you will have a nominal, laminar or transitional flow.

The definition for laminar and turbulent flow is:

Laminar flow	The fluid elements move parallel to each other in distinct paths. In all layers the velocity (fluid elements) moves tangentially to the main flow.
Nominal laminar	Small velocity fluctuations are added to laminar flow. This flow is characterized by small velocity disturbances.
Transitional flow	is laminar flow with spatial and temporal velocity disturbances (fluctuations), which decreases relatively quickly distal to the local flow disturbance. It is a flow between laminar and turbulent, where flow disturbances disappear over time.
Turbulent flow	Three-dimensional, spatial and temporal velocity fluctuations are superimposed on the main flow direction. The flow becomes irregular and chaotic.

A fully developed laminar profile creates a parabolic velocity profile (1) and a fully turbulent flow creates a very flat velocity profile (2). The flow behavior can be calculated with a dimensionless parameter called Reynolds number (Re-number). The Re-number can be calculated with the average velocity over the cross section of the vessel, the diameter and the kinematics viscosity. $Re = (u \cdot d/\nu)$ = (Fig. 1)

For pulsatile flow the Reynolds number should be calculated with a flow rate over one pulse cycle

$$u = V/A \rightarrow Re = \frac{4V \cdot d}{\Pi d^2 \upsilon} = \frac{4V}{\Pi d \pi}$$

Normally, you will never find Reynolds numbers higher than 2300 in blood vessels using the above definition. The entrance length is too short and the pulse wave cannot develop into a turbulent flow.

The non-Newtonian flow behavior of blood can be neglected in straight pipes because the profile is only 3–4% different compared to a fully developed paraboloid in a straight pipe (Fig. 1 right, white arrow).

Methods and models

The influence of the bifurcation angle and the stenosis degree were studied.

We used 1:1 true-to scale, elastic silicon rubber models with a compliance similar to that of the arterial wall. This special technique was described in Biorheology 23, 1986.

The surface in the model reproduces the biological vessel surface. The carotid artery models were installed in a physiologically accurate circulatory system.

The fluid was a polyacrylamid mixture and a water solution which shows a flow behavior similar to that of human blood. Only the thixotropic flow behavior could not be simulated (that means the coagulation of blood). The fluid is transparent and has the same refraction index as the model wall. This is important for the laser measurements. The laser light will not be absorbed and the laser beam is not deflected. Measurements were done with 3D-LDA fiber optic system (DANTEC) in a physiological healthy carotid artery

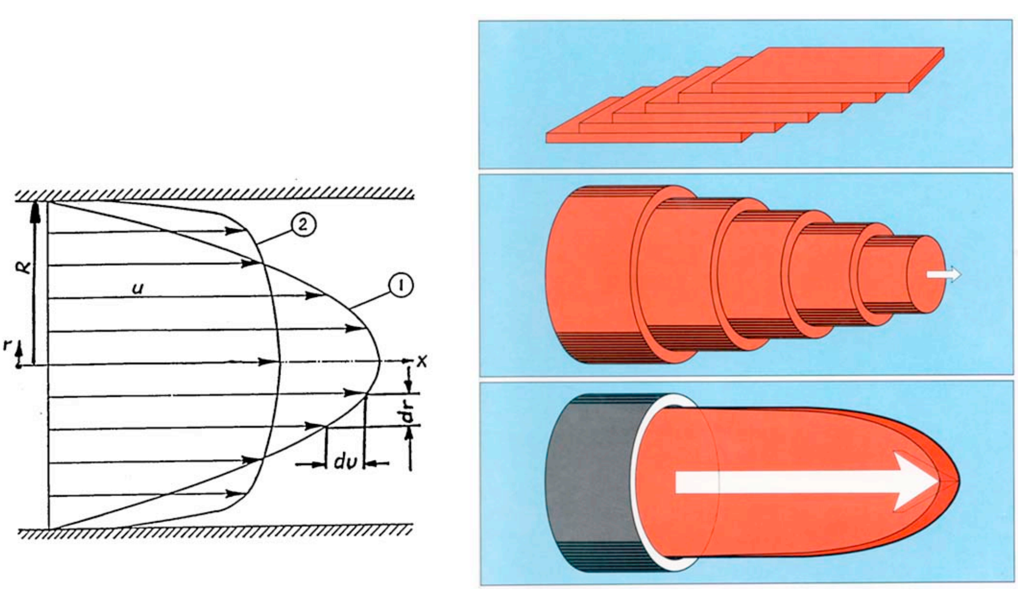

Figure 1 Velocity profile of a laminar and turbulent flow.