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Effect of continuous arterial blood flow in patients with rotary cardiac assist device on the washout of a stenosis wake in the carotid bifurcation: A computer simulation study

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Abstract

In recipients of rotary blood pumps for cardiac assist, the pulsatility of arterial flow is considerably diminished. This influences the shear stress patterns and streamlines in the arterial bed, with potential influence on washout and plaque growth. These effects may be aggravated in the recirculation area of stenoses, and therefore, exclude patients with atherosclerosis from the therapy with these devices.

A numerical study was performed for the human carotid artery bifurcation with the assumption of a massive stenosis (75% reduction of cross-section area) in the carotid bulb. Four different flow time patterns (no support to full pump support) were applied. Flow patterns and particle residence time within the recirculation region were calculated, once within the relevant volume behind the stenosis and and once within a small region directly at the posterior heel of the stenosis.

The flow patterns showed a considerable radial vorticity behind the stenosis. Mean particle residence time in the whole recirculation region was 15% less for high pump support (nearly continuous flow) compared to the natural flow pattern (0.19 s compared to 0.22 s), and nearly identical for the small heel region (0.28 to 0.27 s).

The flow simulation demonstrates, that even in the case of a pre-existing stenosis, the local effects of continuous flow on particle residence times are rather minimal (as was shown previously for intact arterial geometries). Therefore, from the point of macroscopic flow field analysis, continuous flow should not enhance the thromboembolic risk in ventricular assist device recipients.

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1. Introduction

For several years, fully implantable rotary blood pumps have been clinically available for extended cardiac assist and even used for destination therapy for lifelong circulatory support (Wieselthaler et al., 2000; Frazier et al., 2003; Hetzer et al., 2004). These pumps are used to support the left ventricle and are usually positioned with an inflow cannulation to the apex of the left ventricle and an outflow graft to the ascending or descending aorta.

Compared to conventional pulsatile systems these pumps have several advantages such as smaller size, no necessity of valves and membranes and a rotor as the only moving part.

However, in contrast to the natural heart and pulsatile pumps these pumps generate a continuous blood flow, which is considerably different from the physiological condition. Although this flow pattern is usually not

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absolutely constant because of a modulation by the remaining contractility of the supported left ventricle, it is still far less pulsatile than the natural flow pattern in the aortic root (Potapov et al., 2000; Thohan et al., 2005). Typical values of flow observed in the early postoperative phase are in the range of 3–5 L/min with a pulsatility of $\pm$1–3 L/min, resulting in a systolic/diastolic aortic pressure difference of typically 5–15 mmHg. This flow and pressure pulsatility usually increases after recovery of the decompensated heart, but never regains the physiological pulsatility levels.

Therefore, in the early stage of development, there were considerable objections concerning the long-term compatibility of this modified flow with organ function (Thalmann et al., 2005). The clinical experience with long-term implants, which started in 1998, proved the practicability of these devices, their potential for organ recovery and the rehabilitation of patients to excellent quality of life (Wieselthaler et al., 2001; Hetzer et al., 2004). There is no randomized clinical study between pulsatile and continuous devices available yet, but in anecdotal reports the vascular related risks from pulsatile and continuously working pumps seem to be rather comparable. To improve understanding of the modified vascular flow patterns, a numerical simulation study for the carotid artery has been carried out and it was found, that due to secondary flow effects in the radial plane even critical regions such as the carotid bulb does not trap the blood: a spiral washout ensures that particle residence time gets only 20% extended under full pump support (Schima et al., 2003).

However, this first numerical study did not address the important clinical question, if such washout would also exist in arteries with already existing atherosclerotic lesions. A missing washout downstream from stenotic obstructions would probably exclude patients with preoperative atherosclerosis for the application of rotary pumps.

Therefore, we investigated the effects of continuous flow on a carotid artery with severe stenosis. In the following, the results for local flow, shear and washout are presented in their dependency of changing pulsatility for various degrees of continuous pump support.

2. Methods

2.1. Geometry and flow model

The computational model of the bifurcation with a 75% stenosed internal carotid artery (ICA) (Fig. 1) was developed on the basis of a lumen cast of a healthy anatomically realistic model. The preparation and digitalization of the healthy cast were carried out by D. Liepsch, FH Munich. The development of a smooth computational model surface and the finite element grid generation were explained by Karner et al. (1999). The stenosis itself was designed from clinical experiences from both a vascular and a cardiac surgeon, with a 50% reduction in diameter and a 75% reduction in cross-sectional area. Deliberately, it was made intentionally symmetric with respect to the bifurcation plane, to avoid benefits from enhanced spiral washout which was already known from a previous study to play a major role (Schima et al., 2003). The development of the stenosed geometric model used a deformation scheme that can be divided into two steps. In the first step the surface of the initial geometric configuration was contracted in order to generate a smooth stenosis in the ICA. In the next step, the finite-element grid was deformed. This was done by a mapping of the original carotid bifurcation to the stenosed one by solving the 3D-Laplace equation for the deformation of the finite-element grid (Hughes et al., 1981; Prosi et al., 2004).

The time course of mean carotid arterial flow, as it occurs in healthy individuals, was taken from literature (Osenberg, 1991). Data from carotid flow in patients with ventricular assist device were obtained from patients with MicroMed-DeBakey VAD. They were compared with results of a computer simulation at different assist levels from an already available computer model (Vollkron et al., 2002) to get consistent data for different levels of support. The patterns were adjusted to achieve the same mean flow values for all four curve types (Fig. 2). The mean flow rate in the common carotid artery (CCA) over the pulse cycle was 5.5 ml/s shown as horizontal line in Fig. 2. Blood is a non-Newtonian fluid, where the viscosity results largely from the formation of red blood cell rouleaux. At lower shear rates the aggregation of red blood cells causes an increase in viscosity, application of higher shear stress causes disaggregation. While the breakdown of rouleaux happens almost instantaneously the time for reaggregation is mainly determined by the shear rate (Sharp et al., 1996). Due to the relatively high shear rate during the pulse cycle and due to the fact that the pulse period is about ten times shorter than the aggregation time, red blood cells can hardly form rouleaux in large and medium sized arteries, and the Newtonian simplification of blood rheology in larger arteries seems to be justified, applying an appropriate constant reference viscosity. In this study the blood was modelled as a Newtonian fluid with an apparent viscosity $\mu = 3.5\,\mathrm{mPa\,s}$ and a constant density of $\rho = 1050\,\mathrm{kg/m^3}$. The effects of shear thinning and of viscoelasticity in large artery flow have been analysed by Gijsen et al. (1999), Leuprecht and Perktold (2001). A comparison of non-Newtonian inelastic and Newtonian results of flow in a carotid artery bifurcation model has been presented by Perktold et al. (1991).

2.2. Mathematical model

To describe the blood flow the mathematical model employs the time-dependent, three-dimensional, incompressible Navier–Stokes equations for Newtonian fluids. Applying summation convention the governing flow equations for this case are

$$\rho\left(\frac{\partial u_i}{\partial t} + u_j \frac{\partial u_i}{\partial x_j}\right) - \frac{\partial}{\partial x_j}\left(\mu\left(\frac{\partial u_i}{\partial x_j} + \frac{\partial u_j}{\partial x_i}\right)\right) + \frac{\partial p}{\partial x_i} = 0,$$
$$\frac{\partial u_j}{\partial x_j} = 0, \quad i, j = 1,\ 2,\ 3, \tag{1}$$

where u_j, $j = 1, 2, 3$, are the components of the flow velocity and p is the pressure. The system of Eq. (1) must be provided with suitable boundary and initial conditions. At the flow entrance Womersley profiles corresponding to the velocity pulse wave form in the CCA are prescribed as Dirichlet boundary conditions. This assumption is justified due to the length of the common carotid upstream from the computational domain. At the rigid wall, the no-slip condition is applied. At an outflow boundary sufficiently far downstream the region of interest, the condition describing zero surface traction often can be assumed:

$$\left(-p\delta_{ij} + \mu\left(\frac{\partial u_i}{\partial x_j} + \frac{\partial u_j}{\partial x_i}\right)\right)n_j = 0, \quad i, j = 1, 2, 3, \tag{2}$$

where n_j, $j = 1,2,3$, denote the components of the outward pointing normal unit vector at the outflow boundary and δ_{ij} is the Kronecker symbol. In the case of two existing outflow boundaries (internal and external carotid outlet) the flow division ratio was prescribed. Based on the outflow diameter of the two vessels a constant 70%:30% flow partition was assumed in the calculations. This flow partition and also the pulse wave forms were taken from the healthy case, justified by the the fact that the presented stenosis can still be considered moderate (Mansour et al., 1999) and for comparability with the unstenosed condition (Schima et al.,

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