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MRV-validated numerical flow analysis of thrombotic potential of coronary stent designs

Abstract: The formation of a thrombus is associated with dramatic consequences for the patient, such as increased risks of neurologic events and myocardial infarction or even death. A pathologically altered blood flow is associated with these complications. Consequently, a consistent fluid-mechanical analysis of implants, such as coronary stents, must be carried out. Computational fluid dynamics (CFD) is an important *in silico* tool for the analysis of different stent designs. Using three different generic stent designs (closed-cell, open-cell and helical), CFD could be performed with the OpenFOAM software package. The stents were implemented in a vascular model having a fully developed Hagen-Poiseuille velocity profile ($u_{\text{mean}} = 0.6 \text{ m/s}$) as inlet condition. In combination with the dynamic viscosity of the Newtonian test fluid of $6.04 \cdot 10^{-5} \text{ m}^2/\text{s}$, Reynolds numbers up to 460 were achieved. Spatially high-resolved velocity fields from measurements in the magnetic resonance tomograph (magnetic resonance velocimetry, MRV) were available for validation. The velocity field was compared in selected cross sections and longitudinal sections. The difference of the main flow proximal and distal to the stent models were below 6 %. In addition, a similar flow topology could be quantified using the Q-criterion. Due to the very good agreement of the numerical results with the MRV-measurements, the numerical method has been applied to further analysis of stent designs regarding to time average wall shear stress (TAWSS) distribution on the luminal vessel surface (surface area with TAWSS < 0.4 Pa was related to overall vessel surface) under pulsatile conditions. Although all stent designs

have the same square cross-section, a large influence of the stent design on WSS distribution could be observed (closed-cell vs. helical = -50.2 %; open-cell vs. helical = -38.5 %). By using validated CFD it was possible to quantify the hemodynamic benefit of helical stent design in terms of thrombosis potential.

Keywords: CFD, OpenFOAM, stent, magnetic resonance velocimetry, wall shear stress

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1 Introduction

The most common cause of death in Western industrial countries is cardiovascular disease, of which coronary heart disease has the most unfavorable prognosis [1].

The most often used therapy today is the minimally invasive coronary catheter intervention (PTCI), which dilates the stenosed vessel. The coronary vessel can be kept open by means of stent implantation.

Despite the predominantly good properties of stents, complications can also occur after implantation. Mainly restenosis and thrombus are induced as a result of damage to coronary endothelial cells by the stent [2]. The release of drugs through the stent (drug-eluting stent) can prevent these complications to a certain level. However, late stent thrombosis has the highest clinical impact because of the dramatic implications such as stroke and heart failure [2,3].

The stent struts change the blood flow resulting in pathological hemodynamics such as low velocities and consequently low wall shear stresses (WSS) [4]. So the WSS allows estimating the hemodynamic quality of a stent design. Computational fluid dynamics (CFD) gives the opportunity to design stents more efficiently by analyzing many different parameter configurations quickly and cost-effectively. For example, many research groups found that the strut height, which extends into the lumen, is a key parameter for thrombus formation [5-7].

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This is particularly important with regard to recent bioresorbable stents, as a higher strut height is caused by structural-mechanical aspects [3,8]. The intended deflection of blood flows through the strut design promises a targeted improvement of the stent design due to low thrombosis risk. In addition to CFD simulation, the performance of experiments is indispensable for validation in order to prove the plausibility of a numerical model. Magnetic resonance imaging (MRI) has been used successfully in medicine for decades to identify tissue changes. But it can also be used to measure velocities without contact or even optical access, which is needed for Particle Image Velocimetry or laser Doppler anemometry. In contrast to the imaging method, this velocity measurement by means of MRI is called magnetic resonance velocimetry (MRV) [9,10]. Within this work we validated our CFD simulations based on MRV measurements of three generic stent models. CFD further offers the possibility to observe quantities like WSS. So the hemodynamic effectiveness of the stent models with regard to the prevention of thrombosis can be evaluated using the WSS distribution.

2 Materials and methods

2.1 Stent design

Three different generic stent designs, after Stiehm et al. (2013) [11], were analyzed (A: closed cell, B: open cell and C: helical pattern stent), see Figure 1. The stents have a strut thickness as well as width of $150\ \mu\text{m}$ and an angle of 38° between the struts and the main flow direction. The stent models (diameter 3 mm, length 10.5 mm) were scaled by a factor of 16.67 for the MRV measurement to ensure Reynolds-matching. Thus, the stents are located in a scaled vessel model with a diameter of 45 mm and 50 mm at the dilated vessel segment.

Figure 1: three generic stent designs A: closed cell, B: open cell and C: helical stent design

2.2 Computational fluid dynamics

Based on the experimental MRV studies of the flow through the stents, the experimental situation was modelled using a numerical finite volume model.

The vessel model was discretized by means of hexahedral elements arranged in an O-Grid configuration. Then the stent geometry was integrated into the vessel model using the OpenFOAM meshing tool snappyHexMesh (OpenCFD Ltd. (ESI Group), Bracknell, UK).

Figure 2: Magnified view of block-structured vessel model including the stent, meshed using snappyHexMesh

In order to achieve a Reynolds similarity to physiological conditions in coronary vessels, the viscosity and the inlet condition for the MRV measurement were adapted to the enlarged stent model. In the MRV experiment, a Reynolds number of approximately $\text{Re} \approx 460$ was achieved. The Newtonian test fluid used in MRV study was a water-glycerol solution with a dynamic viscosity of $0.073\ \text{Pa}\cdot\text{s}$ and density of $1,210\ \text{kg/m}^3$. The flow was simulated under steady state condition according to the MRV experiment as well as pulsating condition for further flow analyses. A fully developed Hagen-Poiseuille profile with an average velocity of $u_{\text{mean}} = 0.6\ \text{m/s}$ as inlet condition was realized. The pulsatile waveform was taken from [12] and scaled according to the wanted Reynolds number, as seen in Figure 3.

Figure 3: Inlet boundary conditions, steady state and waveform of left coronary arteries (12)

The pressure at the inlet was defined according to a zero gradient condition. At the outlet a zero-gradient condition for the velocity and a fixed pressure of 0 Pa was applied. A no-slip condition for velocity and a zero-gradient for pressure were used at the vessel as well as stent surface. The transient Navier-Stokes equation was solved with the simulation package OpenFOAM-5.0 on high performance computing cluster at Rostock University. Small Reynolds numbers characterizes the investigated flow, therefore no turbulence model was used.

A convergence study was performed according to the wall shear stress on the vessel wall. Convergence of the numerical results occurred at a number of cells above 1.8 million with a relative deviation to even finer meshes of less than 1 %.

Based on the numerical results the thrombosis risk of the three stent designs can be examined. Thrombus formation occurs in areas with low flow. This also correlates with low wall shear stresses. Malek et al. (1999) has defined a threshold wall shear stress of 0.4 Pa [4]. Therefore, all vessel surfaces with $WSS < 0.4$ Pa are treated as thrombotic critical areas (A_{krit}).

3 Results and discussion

The numerical outcomes at the inlet (22.5 mm upstream of the stent) are in accordance with the theoretical Hagen-Poiseuille profile ($u_{mean} = 0.6$ m/s). With a maximum relative deviation of less than 1.1 % and an average relative deviation of 0.37 %, a fully developed Hagen-Poiseuille profile can be assumed at the inlet. Also the profile of the MRV measurements at the inlet shows only minor deviations from the theoretical profile with mean relative deviation less than 6 %. An equal velocity profile was simulated and measured at the outlet (relative deviation < 2 %).

Figure 4: Flow profile at the inlet and outlet of the vessel model, data from MRV measurement (12) and CFD

By analyzing the Q-criterion, which describes the balance between shear and rotation, vortex structures were detected. At $Q > 0$ s⁻² rotation dominates, which together with a local pressure minimum may be an indicator for vortices.

Figure 5: Q criterion in various cross-sections (A and B) of the helical stent, results from MRV measurement (13) and CFD

Figure 5 depicts the calculated Q-criterion in the cross-section of the helix stent, as shown in the sketch (cross-section A and B) based on numerical simulation as well as MRV measurement. In the main flow the Q-values are approximately zero, and therefore no vortices can be assumed. In contrast, high Q-values can be detected in the vicinity of the stent struts. Here, the largest values occur due to a combination of deflection and overflow of the struts. The experimentally determined Q-criterion values correspond qualitatively and quantitatively well with the numerical simulation. Due to the high agreement between the MRV measurement and the simulation of the steady state flow, further numerical flow analyses under pulsatile condition were conducted. Here, the stent designs were compared using the time-averaged wall shear stress (TAWSS).

In Figure 6, the dependence of WSS on the time-varying inflow profile for the open-cell stent design is demonstrated representatively. Red colored areas correspond to A_{krit} .

Figure 6: WSS distribution on the vessel wall with an open-cell stent design in one cycle of the left coronary arteries ($t_1 = 0.25$ s; $t_2 = 0.5$ s), with low WSS (< 0.4 Pa) in red

A_{krit} was analyzed for all stent models at various time steps and was related to the uncovered surface of the vessel, see Figure 7.

Because all stent designs have identical cross-sectional areas, the influence of strut pattern on the WSS distribution can clearly be seen. A_{krit} of the helical design is reduced by 50.2 % and 38.5 % compared to the closed-cell and open-cell design, respectively. This suggests that the selective deflection of the main blood flow by the stent struts improves the WSS distribution. From the hemodynamic point of view, the advantage of a helical design in terms of thrombosis potential could be shown.

Figure 7: Comparison of critical vessel area of the different generic stent designs

4 Conclusion

In this study we have used a numerical model to simulate the conditions of the experimental MRV study of three generic stent designs. Thus the model could be validated on the basis of the velocity profiles and the Q-criterion. It could be demonstrated, that a selective deflection of the blood flow, as shown for the helical stent design, can drastically reduce

pathologically low WSS (< 0.4 Pa). As a result, it is expected that the thrombotic risk can be reduced by the design of the stent itself. Therefore, hemodynamic improved stent design should be considered for future stent development. In particular, bioresorbable stents, characterized by bulky struts, could benefit from hemodynamically driven design improvements.

Author Statement

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