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Numerical simulation and in vitro examination of the flow behaviour within coronary stents

Abstract: This paper discusses the influence of different design parameters of stents by mathematical flow simulations and flow measurements using micro-particle image velocimetry (micro-PIV). A stent strut may cause recirculation areas, which are considered to be the source of thrombosis and the process of in-stent restenosis. The simulations showed that a reduced strut height and a chamfering of the struts reduce these recirculation zones. The numerically determined results were compared with experimental investigations. For this purpose metallic stent structures were transferred into transparent channel systems made of PDMS. The experimental investigations confirm the results of numerical simulations.

Keywords: Stents, Thrombosis, In-Stent-Restenosis, mathematical flow simulations, micro-PIV, PDMS.

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1 Introduction

The cardiovascular diseases and especially the coronary heart disease are among the most frequent causes of death in developed countries today [1]. The causes are mostly arteriosclerotic changes in the coronary blood vessels. Blood lipids, connective tissue and calcium are deposited on the inside of the vessels, which leads to a partial or complete occlusion of the vessel, the stenosis [2].
As a treatment method a stent is usually inserted into the affected vessel. Due to the mechanical supporting effect, the vessel takes on the original diameter of the lumen, which enables sufficient blood circulation again. In addition to the positive effects, clinical complications such as thrombosis or in-stent restenosis may still occur. The cause lies in the changed flow conditions after stent implantation. A stent changes the geometric conditions in the vessel so that the stent strut acts as a flow barrier. Recirculation areas can occur in front of or behind the stent struts. These recirculation areas are characterized by low velocities and low wall shear stresses (τ_w < 0.5 Pa). In these areas thrombocytes can remain behind and attach themselves to the vessel wall thereby promoting the process of in-stent restenosis [3]. For that reason it is important to evaluate the stent design to minimize the disturbing influence of the stent structure for future stent designs.

The aim of this study is therefore to investigate the influence of different design parameters of stents on the flow, such as the strut width, the strut height, an edge rounding or chamfering of the struts. For this purpose, different strut designs are compared to make prognosis about the strength of the recirculation areas in order to optimize them in the future.

In the first step, mathematical flow simulations were performed. In order to validate the numerical results, experimental investigations are carried out using micro-particle image velocimetry (micro-PIV). This is a non-contact measurement method in which particles are added to the fluid to visualize the flows within the stent and thus around the stent struts. This allows the reconstruction the velocity field.

2 Methods and Experiments

For the mathematical flow simulations, a straight rigid tube was modelled as a replica of a coronary artery. This is permissible because the effects of the elasticity in the vessels are very small [4]. The diameter of the pipe was D = 3 mm and the length was 6 times the diameter. As a basic model of a stent, a wave structure similar to a real stent was designed (see Figure 1).

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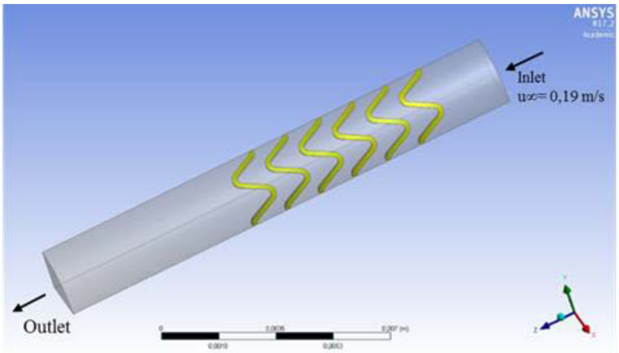

Figure 1: 3D model of the computational domain.

The stent struts have a square cross-section of 150 µm × 150 µm. The inlet-boundary condition was a fully developed velocity profile with an average velocity of u_∞= 0.19 m/s, which corresponds to a representative Reynolds-number at the inlet of 175. This velocity is achieved at rest in coronary arteries and corresponds to the mean velocity over the complete cycle [5].

In order to model blood as realistically as possible, the viscosity change was taken into account by the Carreau model (see eq. 1).

$$\boldsymbol{\eta} = \boldsymbol{\eta}_\infty + (\boldsymbol{\eta}_0 - \boldsymbol{\eta}_\infty)[\mathbf{1} + (\boldsymbol{\lambda}\dot{\boldsymbol{\gamma}})^2]^{\frac{n-1}{2}} \tag{1}$$

For the Carreau model the following parameters were selected [6]: η_∞=0.00345 kg/m-s; η_0=0.25 kg/m-s; λ=25 s; n=0.25. This model fits well to real values for blood [7].

In addition to the classical square strut cross-section, edge fillets (0 to 75 µm), various strut heights (150 µm to 100 µm) and a chamfer of 10° were investigated.

In order to compare the numerical results later with experimental investigations, the PIV method was used. Therefore a micro-PIV setup using fluorescence microscopy was used. In this technique fluorescent particles are added to the flow. The excitation takes place via a LED light source which is coupled into the illumination beam path of the microscope. The stimulating light is directed into the objective via a dichroic mirror and focused on the measuring object. The emitted fluorescent light is then passed through an emission filter and can be registered by the connected CCD camera.

Two images are taken shortly after each other, which are divided into small areas (query areas) for evaluation. From the time difference between the images, the displacement of the individual pixels and the magnification of the microscope objective, a velocity vector in the measuring plane can be determined for each scanning point via a cross correlation function. In order for the particles to be well tracked within the flow, it is necessary that both the entire channel system including the structure to be investigated and the transport liquid have to be transparent.

Therefore channels systems of transparent PDMS were produced. In addition, a transparent 40% mixture of glycerine and water was used as the test fluid. By adding xanthan gum as thickening agent the rheological behaviour of the liquid was given similar to that of blood. Furthermore care was taken to ensure that the viscosity of the liquid corresponds well with the Carreau model used in the mathematical flow simulations.

Due to the curved inner surface it is necessary that the test liquid and the material of the vessel have the same refractive index, otherwise distortions and refractive errors would occur. By adding sodium iodide, a refractive index of n = 1.407 was set for the test liquid at room temperature, so that it was adapted to the PDMS material of the channel.

For the first PIV measurements, transparent channel systems were made from PDMS with simplified stent structures. Laser structuring was used to produce mould inserts from metal rods to be moulded using silicone. The classical square stent structure and a second optimized structure (supported by the simulations) were produced (see figure 2). The ring structures were developed because they are easy to produce and the velocity distribution in the middle plane was well in line with that of the wave structure in the middle plane in simulation calculations.

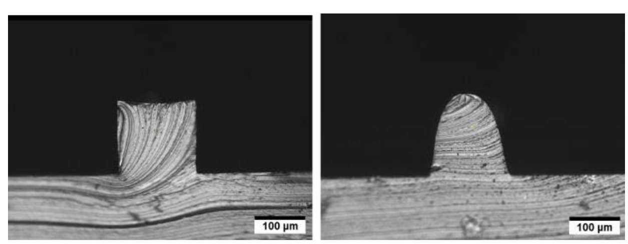

Figure 2: Micrograph of the channels formed in PDMS showing the different strut structures.

However, in order to analyse the influence of the entire stent structure a procedure was developed to convert existing stent designs into moulds in order to transfer them reproducibly into channel systems so that they are accessible for the PIV technique. Therefore the stent was opened by balloon dilatation in a tube with a diameter of 3 mm. The remaining cavities were then filled with PDMS. After the curing of the material the stent and the tube could be removed. This resulting negative is then used to cast a channel with integrated stent structures in PDMS. However, it is necessary to silanise the mould and thus hydrophilise the surface of the silicone. Only then is it possible to later separate the two PDMS layers from each other. In order to check the transfer of the original stent into silicone and the quality of the structures, they were characterized by using a scanning electron microscope (see figure 3).

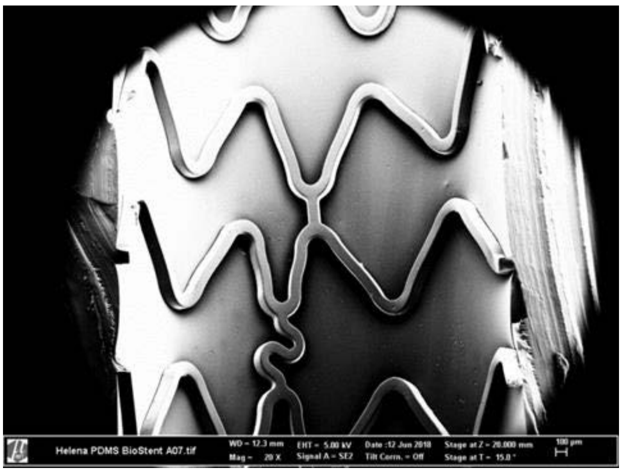

Figure 3: Scanning electron micrograph of the inner surface of the channel formed in PDMS showing the stent structure.

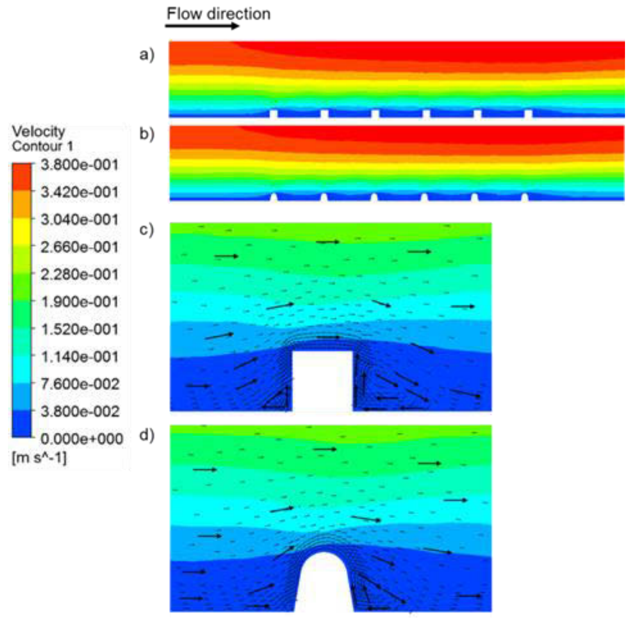

Figure 4: Results of the flow simulation within the stent models. a, b) entire flow area in the middle plane of the stents; c, d) detailed representation of the resulting speed profile around the respective third stent strut. The long black arrows illustrate the formation of the recirculation areas. The colour scale shows the respective speeds.

3 Results

The mathematical flow simulations clearly showed the representation of the secondary flow and thus the recirculation areas around the stent struts. It could be shown that a reduced strut height and a chamfering of the struts reduce these areas. A larger recirculation zone can be seen behind the structure in the downstream area. In addition to the formation of the secondary flow, a slight acceleration in the centre of the flow can also be detected.

Despite the same height of the investigated structures, the optimized strut shape with an edge rounding and additional chamfering already achieved a larger reduction of the recirculation zones (see Figure 4).

The PIV measurements confirm the results already achieved in the simulations (see Figure 5). However, it should be noted that the maximum speed achieved in the upper edge area exceeds that calculated by simulation. This can be explained by the elasticity of the PDMS.

In addition, when subdividing the interrogation areas, care was taken to resolve the recirculation area well, so that they may have become too fine in the upper edge area.

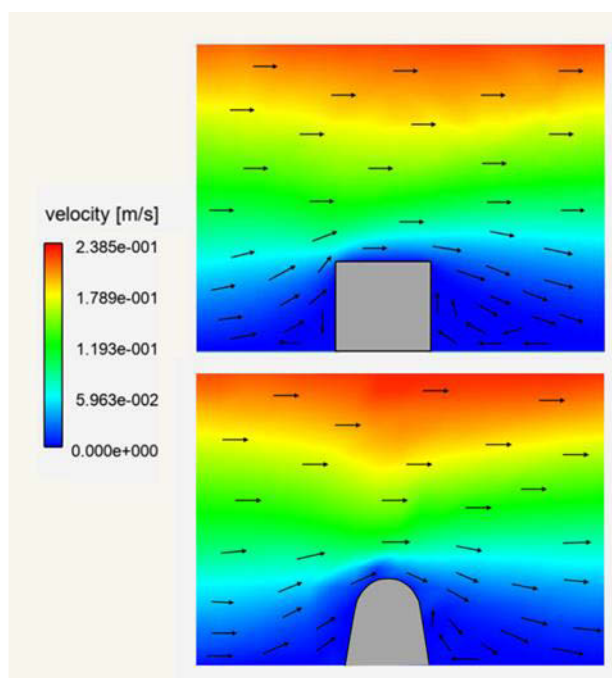

Figure 5: Results of the PIV-measurement. Detailed representation of the resulting speed profile around the top: square stent struts and bottom: optimized struts. The black arrows illustrate the formation of the recirculation areas. The colour scale shows the respective speeds.

In order to assess the wall shear stress (WSS) as a criterion in addition to the pure velocity field, these were determined from the PIV measurements. The velocity gradient was determined from the 2d velocity field. Taking into account the viscosity by the Carreau model, the wall shear stress could be calculated. These values were compared with the calculated WSS values by the flow simulation (see Figure 6).

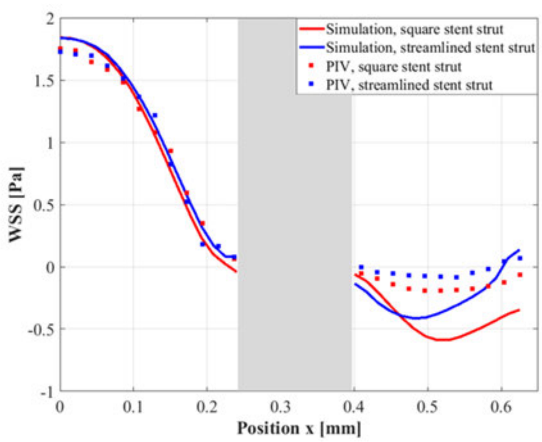

Figure 6: Diagram of the calculated wall shear stresses around the stent struts of the flow simulation (lines) and the PIV measurements (dots). The grey area represents the position of the stent struts.

The graph clearly shows the achievement of a critical wall shear stress of τ_w<0.5 Pa. As expected, these critical ranges are achieved in the recirculation areas. Due to the optimized structure, reductions of the critical areas are possible and representable.

4 Conclusion

The aim of this study was to investigate the influence of different design parameters of stents. Mathematical flow simulations were compared with experimental investigations using micro-PIV. Different stent strut designs were investigated. It could be shown that a reduced strut height and a chamfering of the struts have a positive influence on the flow conditions. In order to compare these numerically determined results with as real models as possible, transparent channel systems were produced from PDMS. Metal stent structures were transferred for this purpose. The PIV-measurements confirmed the numerical results.

5 Outlook

In this study, the PIV-measurements were performed with a stationary inflow with a velocity of u_∞= 0.19 m/s. This velocity is obtained at rest in coronary arteries. However, since the blood flow is not subject to a natural pulsatile velocity, future experimental investigations will be carried out with a pulsatile inflow. Furthermore, the focus of this study was on the optimization of the shape of the stent struts. In the future, a more precise analysis of the entire stent structure will be carried out.

Author Statement

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References

[1] Cardiovascular diseases. World Health Organization. url: https://www.who.int/news-room/fact-sheets/detail/.
[2] Wintermantel, Erich and Ha, Suk-Woo, Ed., Medizintechnik: Life Science Engineering, Springer Berlin Heidelberg, 2009.
[3] J. J. Wentzel, F. J. H. Gijsen, J. C. H. Schuurbiers, A. F. W. van der Steen, P. W. Serruys, The influence of shear stress on in-stent restenosis and thrombosis, Euro Intervention Supplement 4 Supplement C, pp C27–C32, 2008.
[4] Ku DN., Blood flow in arteries, Annu Rev Fluid Mech,29 : pp 399–434, 1997.
[5] Nichols WW, O'Rourke MF, Vlachopoulos C., McDonald's Blood Flow in Arteries: Theoretical, Experimental and Clinical Principles, New York Hodder Arnold, 2011.
[6] Taewon Seo, Levanto G.Schachter and Abdul I. Barakat; Computational Study of Fluid Mechanical Disturbance Induced by Endovascular Stents, Annals of Biomedical Engineering, Vol. 33, No. 4, pp. 444–456, April 2005.
[7] Chien, S., S. Usami, M. Taylor, J. L. Lundberg, and M. I. Gergersem, Effects of hematocrit and plasma proteins on human blood rheology at low shear rates, J. Appl. Physiol. 21: pp 81–87, 1966.