FLOW OPTIMIZATION OF A MICRO-AXIAL BLOOD PUMP

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Abstract: A detailed knowledge of the flow field in a blood pump is indispensable in order to increase the efficiency of the pump and to reduce the shear induced hemolysis. Thus, three different impeller designs were developed and tested by means of digital particle image velocimetry (DPIV). An optimization of the impeller could be achieved by following the concept of turbulent drag reduction for the hub.

Introduction

In recent years the tendency of miniaturization has shown promising results for applications of medical instruments in keyhole-surgery. Small rotary pump systems have been developed for the treatment of patients with cardiac insufficiency as well as for applications in intra- and extra-corporal circulation. Any hemodynamical and fluid mechanical optimization of the pump will extend substantially the period of pump implementation which is of great significance and impact for cardiology, the patients and the health care system. The micro-axial blood pump to be analyzed in this study has been developed by Impella Cardiosystems AG for a temporary support of the left ventricle in acute situations [1]. It has only an overall outer diameter of 4 mm and is placed on a catheter via the arteria femoralis into the left ventricle (Fig. 1). Thus, it offers new approaches in minimal invasive treatment of patients, which can be handled even in cardio labs.

Materials and Methods

Microaxial blood pump

The Impella®acute micro-axial blood pump has an overall outer diameter of 4 mm and is placed on a catheter via the arteria femoralis into the left ventricle. The pump in its initial design circulated a maximum blood flow of 2.4 ltr/min at 50 mmHg and peak rotational speeds of 50000 rpm. Higher flow rates could only be achieved with a dramatic increase of the hemolysis. Therefore it is essential to improve the pump efficiency towards higher flow rates without increasing the hemolysis.

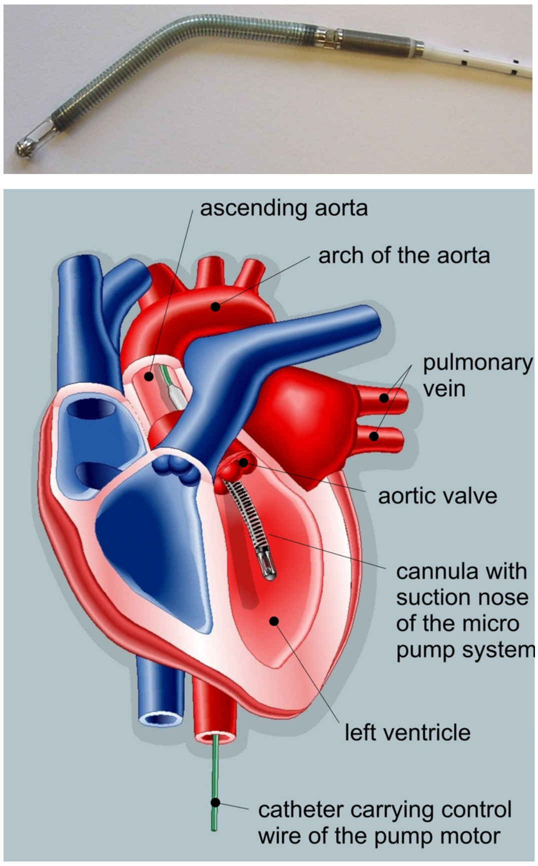

Figure 1: The photograph on top shows the left ventricular micro-axial blood pump (Impella CardioSystems AG, Aachen, Germany) including the motor and the inflow cannula (diameter 4mm); (bottom) Implementation of the impella®acute pump system into the left ventricle

Different profiles of the center-body and the impeller blade were tested to improve the hydraulics of the pump and reduce the blood cell traumatization. Three different types of impeller geometries as shown in Fig. 2 are analyzed. The most promising and feasible improvement could be achieved with a impeller design leaned upon aerodynamic drag reduction experiences.

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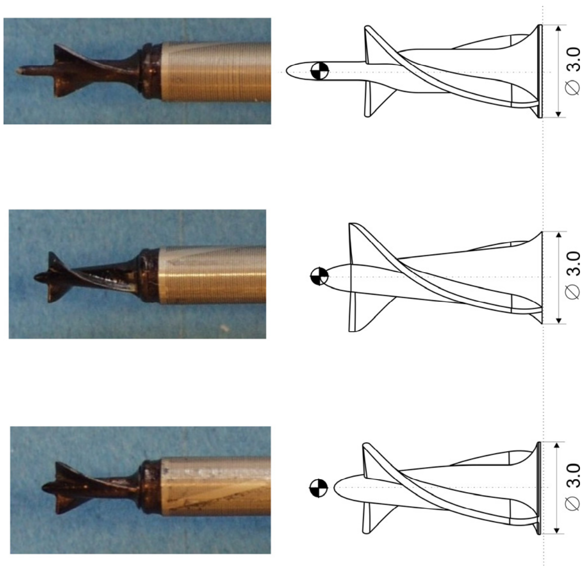

Figure 2: The different geometries of the impeller; (top): Im 569 with elongated forebody; (center): Im 555 corresponding to the basic configuration; (bottom): Im 388 with shortened hub

It is known that the drag of an axisymmetric body can be reduced in some cases by a forebody nose, which is most effective in turbulent flows. The drag reduction is mainly due to the redistribution of the wall shear stress (higher wall shear along the forebody with a small effective area, lower shear stress along the main body with a much larger effective surface area) and the stabilization of the boundary layer in the successive convex parts of the nose as described by [2] (Fig. 3).

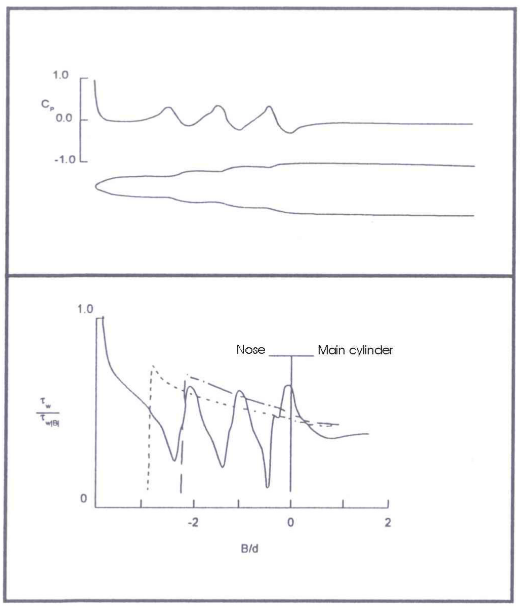

Figure 3: Comparison of wall shear stress profiles along a three-stage nose body compared with equivalent-area (dashed line) and equivalent-volume half-elliptic noses (chain line), [2]

The hydraulic performance curves of the three different impeller are shown in Fig. 4.

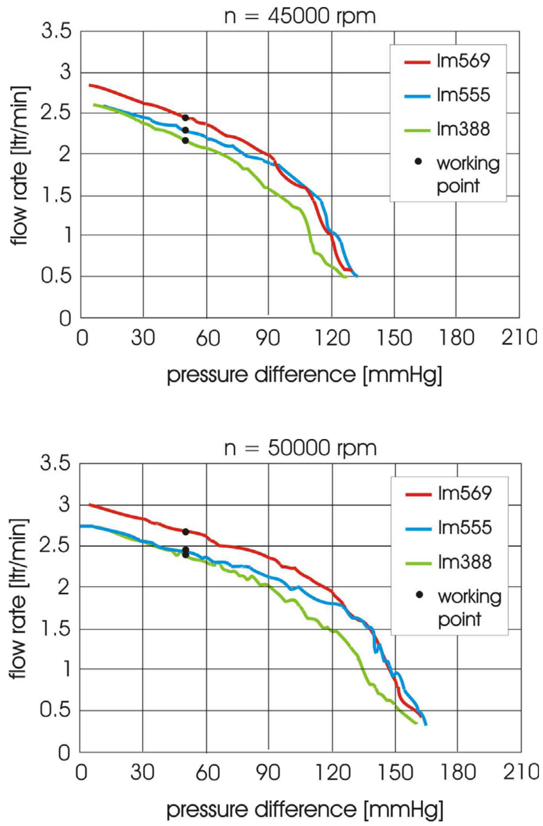

Figure 4: Hydraulic performance curves of the different impeller, demonstrating the improvement of pump efficiency of impeller Im 569 with elongated forebody

Experimental set-up

For the purpose of this study the impeller is integrated into a hydraulic mock loop (Fig. 5).

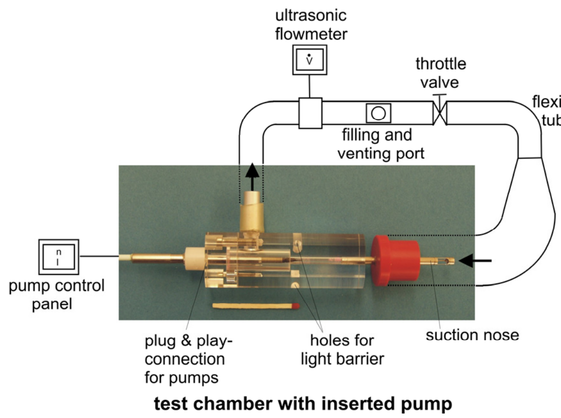

Figure 5: Flow circuit for pump tests

A water/glycerine mixture (35 wt % glycerine) is used as working fluid. Flow rate is controlled by an ultrasound flow meter and the pressure load is regulated by a throttle valve downstream of the pump. The connector of the impeller to the mock circuit allows an easy change of the different impeller types.

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Detailed measurements of the internal flow within the micro-axial pump are carried out using digital particle-image velocimetry (DPIV). This method allows in principle to determine the planar flow field in a selected plane of the flow, which is defined by a light sheet. Therefore, the laser beam of a Nd:Yag pulse laser is expanded by a lens system and redirected from top and bottom into the test chamber. A micro-positioning stage allows to move the test chamber through the light-sheet with high precision. The full optical arrangement can be seen in Fig. 6.

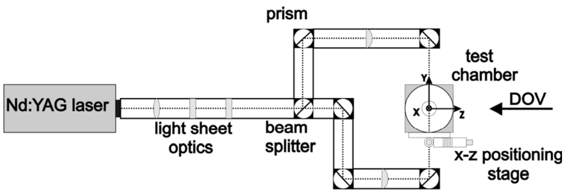

Figure 6: Optical arrangement (DOV - direction of view) [3]

The flow is seeded with particles and recorded by a double-shutter PCO-camera.

Measurements were taken in the axial centerplane and in planes with axial offset. A light barrier allows to synchronize the laser pulse to any desired angular phase position of the impeller. Velocity fields are calculated by cross-correlation procedure with window shifting and window refinement technique. The final results represent averages of 50 DPIV recordings at the same phase position. From these data, vorticity is calculated and velocity profiles are reconstructed at different axial locations in main flow direction.

Results

Velocity flow profiles

The PIV measurements could, for the first time, visualize the axial velocity profiles in these small geometries and for extremely high rotational speeds of 50000 min^{-1}. The velocity flow profiles in a centerplane of the optimized impeller design are shown in Fig. 7. In Fig. 7 (a) the velocity profiles show a fully developed turbulent inflow which corresponds exactly to the condition in the real inflow cannula for this working point. The flow is decelerated near the stagnation point of the hub and in near-wall regions. The outlet region, which simulates the annular flow region downstream in the aorta, experiences a recirculation that influences the outflow in a positive way. The maximum velocities in the outlet region can reach values up to the threefold the mean inflow velocity.

Spanwise vorticity distribution

Fig. 8 shows the evolution of the flow (in steps of 18°) in the impeller region of the optimized pump design. The color indicates the amount of the spanwise vorticity component. The vorticity is normalized with

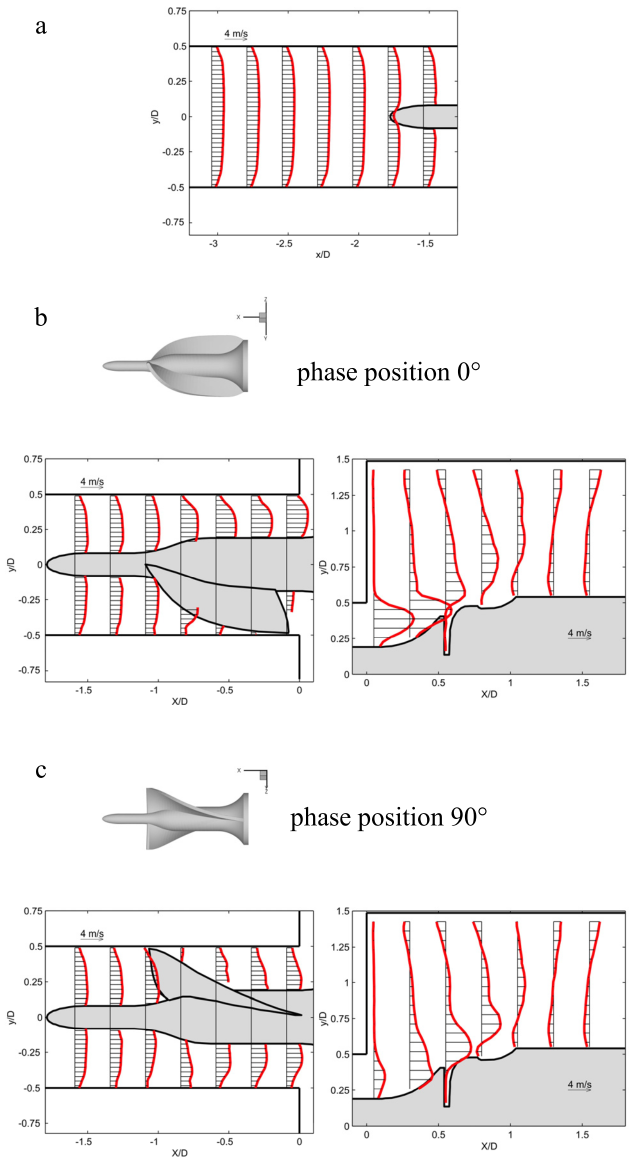

Figure 7: Axial velocity profiles; (a): inflow region, (b): impeller and outlet region at blade phase position 0°, (c): impeller and outlet region at blade phase position 90°.

the angular speed of the impeller $\omega_{impeller}$. It is clearly visible that the upper tip clearance vortex has a stronger presence in the x-y-plane in phase positions 144°, 162° and 0°, whereas the lower tip clearance vortex is strong in phase positions 18°, 36° and 54°.

The axis of the tip clearance vortex is tilted as the vortex moves downstream. At its origin close to the leading edge of the blade the vortex axis is more or less perpendicular to the x-y-plane and becomes parallel to it during its backward drifting. The path of the tip clearance vortex has been reconstructed by interpolating the vortex position data from Fig. 8 and assuming a vortex diameter of 0.1D, D being the diameter of the inflow cannula. The result is shown in Fig. 9 as seen on a co-rotating frame of reference.

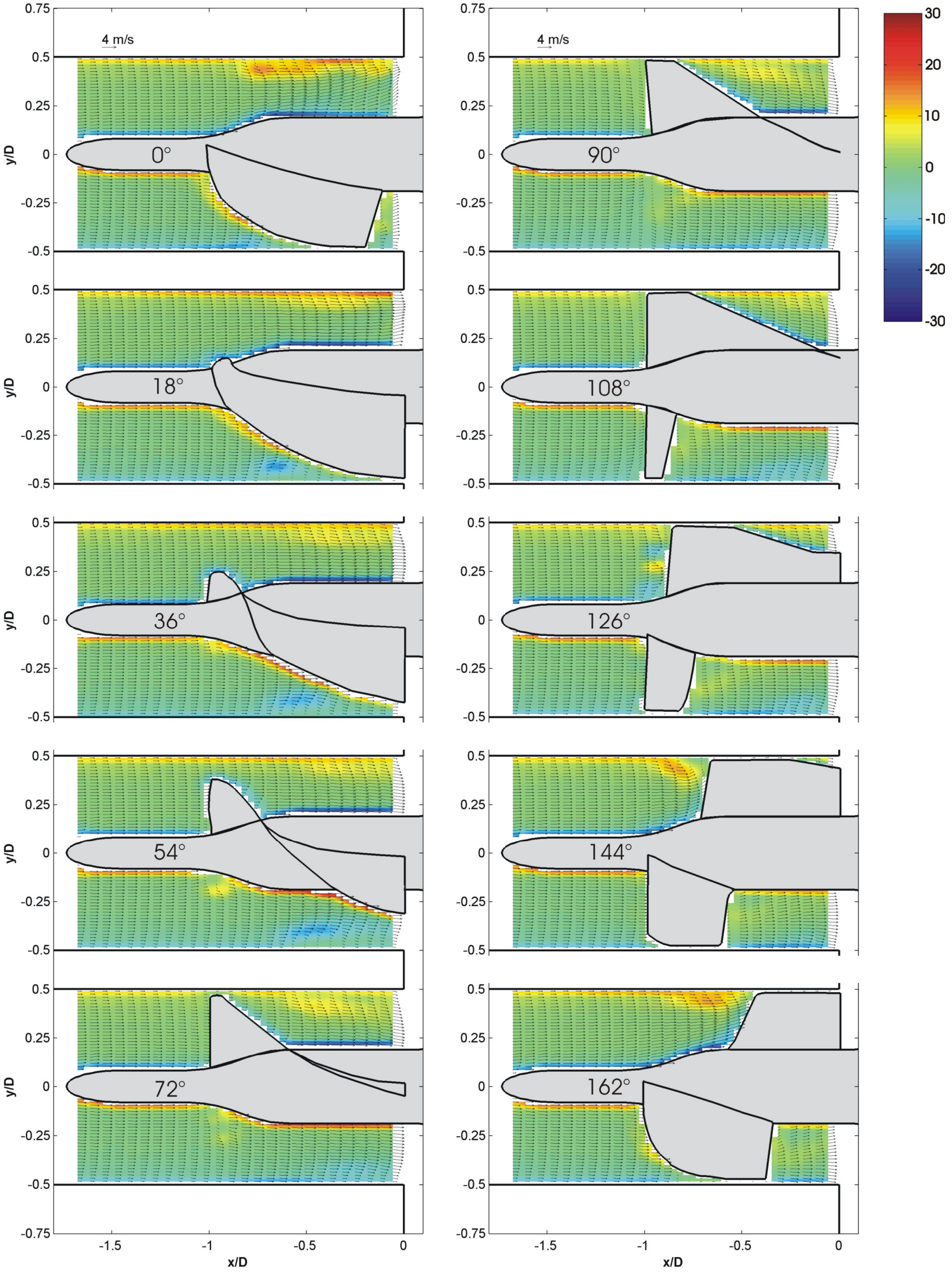

Figure 8: Evolution of the flow structure in the centerplane over one revolution of the impeller in steps of 18°. The color indicates the amount of the spanwise vorticity component

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Fig. 9 points out that the tip clearance vortex first follows the curvature of the blade and then starts to diverge from it. The figure also demonstrates that even at a pressure load of 50 mmHg and a high rotational speed of 50000 min^{-1} the vortex will not be intercepted by the subsequent blade. This 2-blade concept of the impeller will lead to better hemolytic properties than a 3- or 4-blade rotor.

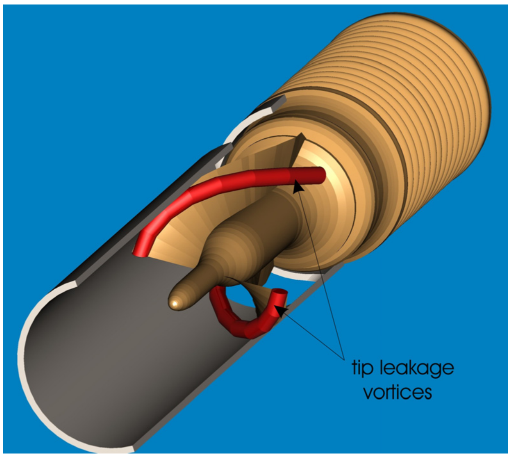

Figure 9: Cut view of the inflow cannula and the pump housing showing the impeller and the path of the tip leakage vortex as seen on a co-rotating frame of reference

Discussion and conclusion

The flow through three different types of rotor geometries was studied by means of the technique of Digital Particle-Image Velocimetry (DPIV). This method allows determining the planar flow field in selected planes of the flow. The DPIV measurements were carried out for various working conditions.

The most beneficial modification could be obtained by changing the nose of the center-body shape in the form of a two-stage nose body by which the flow rate of the pump was increased by 10 percent at the same rotation rate. This improvement allows to operate the pump at reduced rotational speeds of 45000 rpm while producing the same flow rate as the original pump at 50000 rpm, but more important at much lower hemolysis levels.

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