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Outlet Optimization of the Centrifugal Blood Pump

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Abstract: The Computational Fluid Dynamic (CFD) package COSMOS, was applied to simulate the flow through three versions of a centrifugal blood pump modeled with Solid Works. In this study, the 3 dimensional flow patterns through the artificial heart pump was predicted. The simulation was performed to detect Maximum Shear Force (N) through the pump, Averaged Static Pressure (Pa), Averaged Outlet Velocity (m s^{-1}), Pressure Drop (Pa) and Averaged Outlet Total Pressure (Pa). Three versions have been simulated model; the basic of Nikkiso blood pump with a circular outlet, the same pump modified with a rectangular outlet and a modified pump with a semi rectangular outlet. The pumps were simulated at T 37°C; with an impeller rotational speed of 3000 rpm clockwise, the cross sectional area of the outlet window was kept constant at 50 mm^2 for all versions and human blood was identified as working fluid with input/output volume flow rate of 6 L min^{-1}. The simulation results indicate that model (3) with a semi rectangular outlet has a better performance and gave the lowest average of static pressure and the lowest blood hemolytic levels with decrease of about 14.12% from the base pump model (1).

Key words: High shear stress, blood hemolysis, bio-compatibility, centrifugal pump, bio-fluid engineering

INTRODUCTION

Heart disease is the number one killer of adults in the western world (Ridker, 2002). When KOLFF`s group at the Cleveland Clinic first displaced their artificial heart project at the scientific Exhibit of the Chicago Meeting of the American Medical Association in June 1962, they defined an artificial heart as follow: An artificial heart is a mechanical prosthetic heart which completely substitutes for the natural heart, anatomically and physiologically. It will be inserted inside the chest after total resection of an irreparably diseased natural heart (Akutsu, 1975). A Ventricular Assist Device (VAD), or mechanical heart, is a mechanical device used to partially or completely replace the function of a failing heart located in the patient`s chest. It's important role is to support the native heart by pumping blood from the heart left ventricular chamber to the aorta and to all the body. Blood pumps are widely used as bridge to either recovery or transplantation and are even being used as destination therapy for patients with end-stage heart failure (Bunzel *et al.*, 2007). The demand for qualified, bio-compatible artificial circulatory support systems for temporary support or permanent replace an irreparably human heart is well established. This need became apparent to cardiac surgeons in the early 1950s after the clinical application of the first hear lung machine. Due to a shortage of donor hearts for transplants, the demand for artificial heart pumps has lead to worldwide investigation and development of mechanical assisted devices including pneumatic pumps, axial flow pumps and centrifugal pumps. Axial flow pumps, with the advantages of smaller volume, higher efficiency and lighter weight, are widely used in Ventricular Assist Device (VAD) systems (Noon *et al.*, 2000; Macris *et al.*, 1997; Kilic *et al.*, 2000). Some VAD's are intended for short term use, typically for patients recovering from heart attacks or heart surgery, while others are intended for long term use months to years and in some cases for the life of the patient. These long term uses are typically for patients suffering congestive heart failure. VAD's need to be clearly distinguished from artificial hearts, which are designed to completely take over cardiac function and generally require the removal of the patient native heart. Among the available blood pumps the centrifugal blood pump has received the most interest for use as VAD because it permits a compact design and can achieve high blood flow rates. Hemolysis tests and animal experiments have proved that the blood pump has an acceptable but not ideal hemolysis level; thrombus occurred and Computational Fluid Dynamics (CFD) analysis showed regions of reverse flow present in the diffuser, which not only decreases the pump's hydrodynamic efficiency, but also increases its overall potential for blood trauma and thrombosis (Zhang *et al.*, 2008). One of the crucial challenges for current VAD's is blood trauma including

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both hemolysis and thrombosis resulting from high shear stress and stagnation effects on as the blood passes through the VAD. Normal human blood contains Red Blood Cells (RBC`s), which are the most crucial element of the blood, hemolysis is the rupture of the (RBC's), which releases hemoglobin to the blood plasma. It is caused by high shear stress from mechanical pumping of the blood.

Centrifugal fluid pumps are well known in the hydraulic and pneumatic fields. They typically consist of a motor to drive a shaft on which a fluid impeller is mounted. Generally, the fluid inlet or suction port feeds fluid to the hub or the center of the impeller. A number of impeller vanes generally project outward from the hub in spiral paths and are supported between shrouds, which, together with the vanes, constitute the pumping channels. The impeller is encased in a housing that channels the working fluid from the inlet port to the hub, or the inducer, where it is inducted into the pumping channels between the vanes and the shrouds. The centrifugal action of the impeller drives the working fluid outward to a diffuser at the periphery of the impeller disk, where it enters a scroll-shaped volute and from there it is channeled to the discharge port of the pump (Turton, 1993; Cheremisinoff, 1993; Evans, 2000). The most successful VAD's are designed to be more bio-compatible and induce lower shear stress on the blood. A number of CFD studies have investigated the effect of geometry on the flow behavior in the pump. There are many applications where a more compact form of the pump is desired. There are

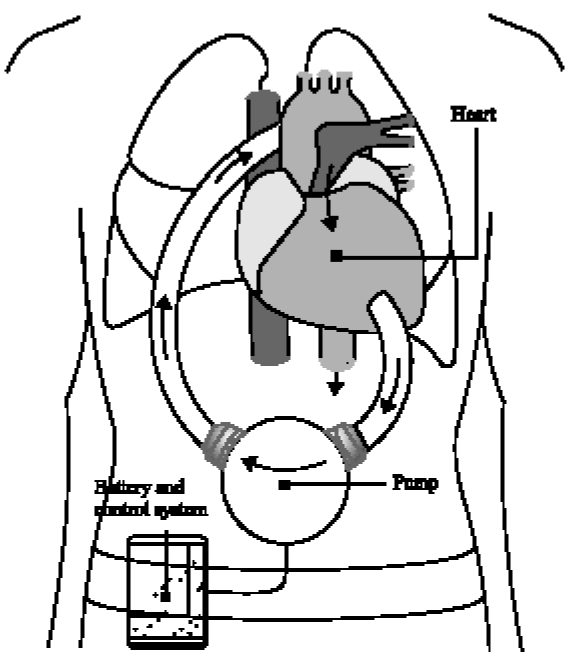

Fig. 1: The artificial heart system components

also, applications where the shaft and the seals can present operational problems. A well-known example is that of the artificial heart pump where dangerous blood clots can form in the areas where the motor shaft enters the pump housing (Miller *et al*., 1990; Kovacs *et al*., 1980; Shen *et al*., 2000).

In this research, we focus on the effect of changing the outlet geometry on the flow behavior, Maximum Shear Force (N), Averaged Static Pressure (Pa), Averaged Outlet Velocity (m s^{-1}), Pressure Drop (Pa) and Averaged Outlet Total Pressure (Pa). Figure 1 shows the artificial heart system main components.

MATERIALS AND METHODS

In this research, we modified the outlet of an existing model of Nikkiso pump. Solid Works and COSMOS flow works were employed for modeling and simulation. The goal was to reduce shear stress on the working fluid and investigate flow behavior through the pump. Three models were designed with same basic parameters and dimensions.

Figure 2 shows the 3D view of the basic simulation model of Nikkiso Blood Pump model.

Three different outlets were used. The structure of a 6 blades pump base model is shown in Fig. 2. The impeller

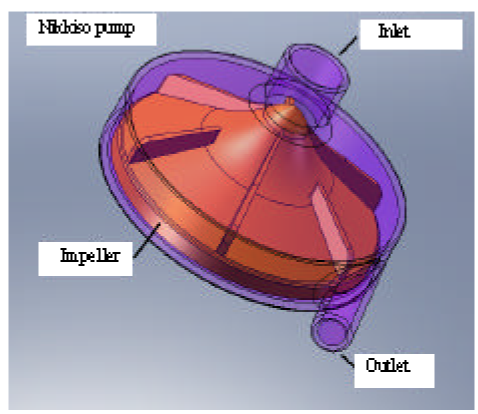

Fig. 2: 3D view of the basic simulation model of Nikkiso Blood Pump

Table 1: The design parameters of the pump model

Model specifications	
Impeller diameter	80 mm
Impeller high	35 mm
Number of blade	6
Rotation speed	3000 rpm
Volume flow rate	6 L min^{-1}
Output cross section area	50 mm^2
Working fluid	Human blood
Temperature	37°C
Rotation direction	Clockwise

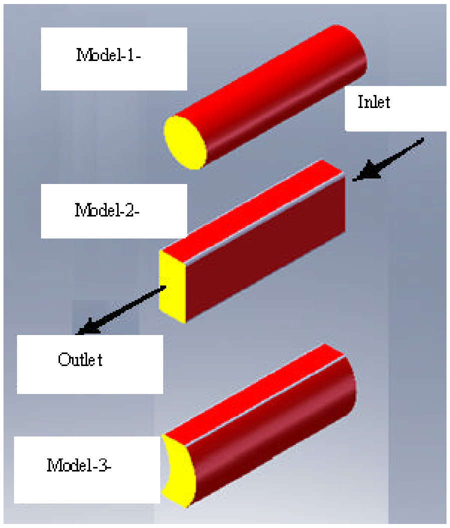

Fig. 3: 3D view of the outlet geometry design. 1): Model circular outlet, 2): Model rectangular outlet and 3): Model semi-rectangular outlet

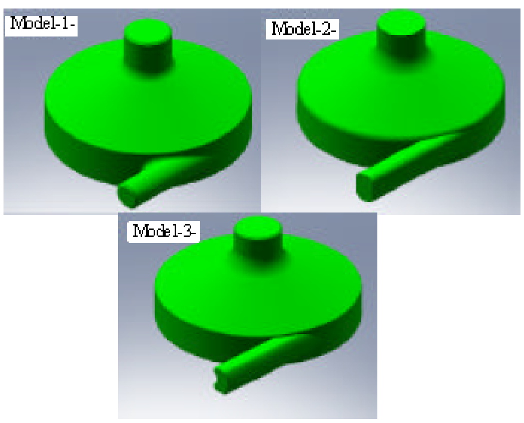

Fig. 4: 3D view of the 3 different simulation models of the pump outlet

has 6 blades they were three completed and 3 half blades. The casing has one tangential discharge circular outlet. The outer diameter of the impeller is 11.97 mm and the hub diameter is 7.978 mm. Radial clearance between the impeller and the housing is 1 mm. The casing inner diameter is 82 mm with a 16 mm diameter inlet, a 7.97 mm diameter outlet and casing thickness is 2 mm.

As shown in the Table 1 the only difference between the models is the outlet geometry. This can be seen clearly in Fig. 3. The cross sectional area of the 3 models has been kept constant at 50 mm^2 the 1st model is the base Nikkiso design with circular outlet, model 2nd has rectangular outlet and model 3rd has a semi-rectangular shaped outlet.

Table 2: The main project definitions procedure and characteristics

COSMOS flow works 2007	Common project definition
Unite system	SI unit (m-kg-s)
Analysis type	Internal flow
Rotation condition	2000 rpm, clockwise
Input boundary condition	Inlet volume flow
Input boundary condition	Outlet environmental pressure
Initiate mesh	2016 Cells
Wall thermal condition	Adiabatic wall
Calculated results	Maximum shear force (N)
	Averaged static pressure (kPa)
	Averaged outlet velocity (m s^{-1})
	Pressure drop (kPa)
	Averaged outlet total pressure (kPa)

Therefore, it can be seen in Fig. 4 the 3D view of 3 different simulation models of the pump outlet.

The Computational Fluid Dynamics (CFD) program COSMOS Flow Works used to calculate the three models performance and to investigate the flow behavior in each case.

Table 2 shows the main project definitions procedure and characteristics.

RESULTS

A cubical mesh was used in all cases; Table 3 gives meshing statistics from the models.

Table 4 shows the estimated values of Maximum Shear Force (N), Averaged Static Pressure (Pa), Averaged Outlet Velocity (m s^{-1}), Pressure Drop (Pa) and Averaged Outlet Total Pressure (Pa) for the 3 pump versions.

As can be seen in the bar Fig. 5a model (1) has the lowest value of maximum shear force (0.305 N) while, model (2) has the highest value of maximum shear force (0.372 N) and the maximum shear forces in both of models (2) and (3) was very similar.

The bar Fig. 5b point out the estimated values of the average outlet velocity for the three pumps versions. Model (3) has the highest value of the average outlet velocity (3.84 m s^{-1}) and model (2) has the lowest value at (2.69 m s^{-1}).

As can be seen in Fig. 5c the model (3) has the highest outlet total pressure (111 kPa) and pressure drop (76 kPa) while, it has the lowest average static pressure (65 kPa). Among the three pump versions model (2) has the medium values of the Averaged Static Pressure (68 kPa), Pressure Drop (73 kPa) and Averaged Outlet Total Pressure (108 kPa). Model (1) has the lowest pressure drop (59 kPa) and the highest static pressure (75 kPa), while its outlet total pressure was (109 kPa) was very similar to its value in the model (2) and (3).

Blood hemolysis has a direct relation with high shear stress and turbulent flow which generate by the pressure on the blood. In the model 3 was the lowest static

Table 3: Meshing results

	Project simulation meshing results		
Model	1	2	3
Total cells	37.352	38.605	39.361
Fluid cells	8.863	9.009	9.038
Solid cells	12.565	13.208	13.576
Partial cells	15.924	16.388	16.747
Irregular cells	0	0	0
Iteration	268	271	546
Initial mesh cells	2.016	2.016	2.016

Table 4: Total evaluation for the three pump versions

Model	1	2	3
Maximum shear force (N)	0.305	0.372	0.360
Averaged static pressure (kPa)	75	68	65
Averaged outlet velocity (m s^{-1})	3.06	2.69	3.84
Pressure drop (kPa)	59	73	76
Averaged outlet total pressure (kPa)	109	108	111
Total evaluation	Bad	Medium	Good

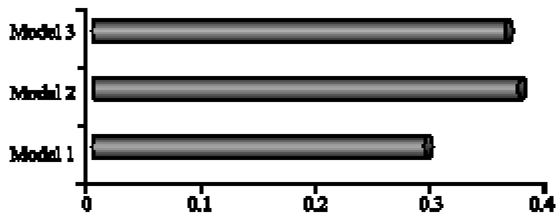

Fig. 5a: Maximum shear forces for the 3 pump versions

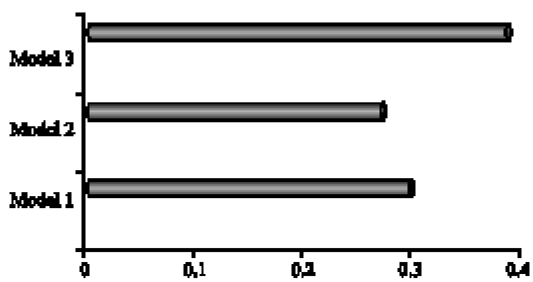

Fig. 5b: The average outlet velocity for the 3 pump versions

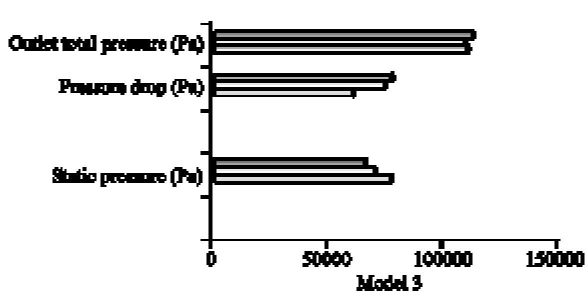

Fig. 5c: The averaged static pressure (Pa), pressure drop (Pa) and averaged outlet total pressure (Pa) for the 3 pump versions

pressure (65 kPa) which means the lowest shear stress and blood hemolytic, while it was (75 kPa) and (68 kPa) in the models 1 and 2, respectively.

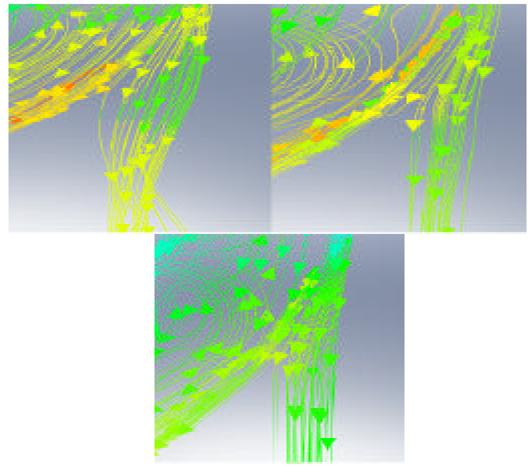

Fig. 6: Top views of the flow profiles

Figure 6 show the flow streamlines in the model 3 the flow behavior seems less turbulence flow than its profiles in models 2 and 1 and the simulation data results indicate that model 3 with a semi rectangular outlet has the best performance and gave the lowest turbulence and static pressure and the lowest blood hemolytic levels with decrease of about 14.12% from the base pump model 1.

Based on the physics the streamlines more likely behave laminar when they run on the plane surfaces this phenomenon applicable for the outlet models 2 and 3, however the curvature surfaces allow to the streamlines to form vortexes and turbulent behavior and this is exactly applicable for the outlet model 1.

DISCUSSION

First of all, this first generation design demonstrate the potential of this system to support the geometry design studies of the ventricular assistant devices to put the final design improvement to achieve low levels of hemolysis.

Secondly, the results and data pointed out that model 3 of semi-rectangular outlet has the best performance and gave significant laminar flow among the three models, but in fact, that's may be real because of the ideal conditions and walls which have been identified during the simulation and this design improvement need to be verified with an experimental data.

In addition, we identified the ideal conditions and properties of the working fluid to simulate the real blood and this fluid behavior needs to be verified by using the real blood in our future experiments.

Finally, this research needs to detect more factors to improve the design such as blade number and geometry, rotational speed, gravity, impeller casing clearance and outlet position.

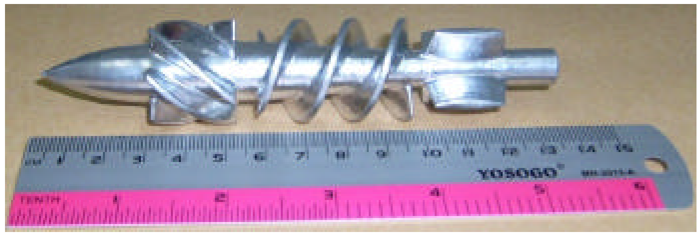

Fig. 7: Shaft bearing axial impeller pump

Fig. 8: Full suspending magnetic bearing axial impeller pump

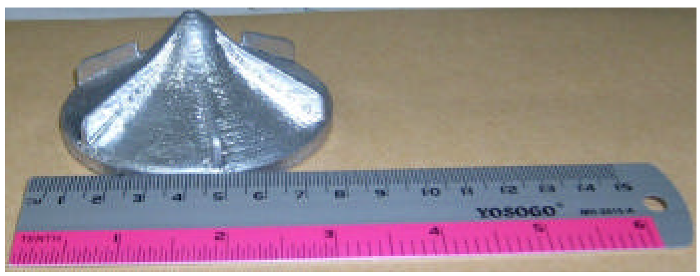

Fig. 9: Centrifugal impeller pump

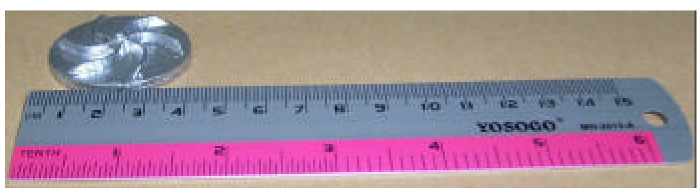

Fig. 10: Pediatric centrifugal impeller pump

Future work: Further research in this area will be focus on the design, prototyping and system fabrication and data verification for both kinds of both the heart axial and centrifugal blood pumps.

Some of the creative components can be seen in Fig. 7-10 have been done. In addition to other kinds of special purpose impellers for pediatric and adult circulatory supports systems.

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