Analysis of Rotary Ventricular Assist Devices using CFD Technique - A Review

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Abstract

With the increasing focus on reducing deaths because of heart ailments, considerable emphasis has been directed towards the development of ventricular assist devices (VADs) that work on the principle of mechanical pumps for supporting an infirm heart. The VADs are used primarily to assist the left ventricle, but their use has been extended for supporting the right ventricle as well as for supporting both ventricles simultaneously. In this connection, computational fluid dynamics (CFD) has evolved as an efficient technique for the design, development, evaluation, and optimization of VADs, enabling the prediction of operational characteristics of VADs as well as detailed global and local flow characteristics for an extensive set of working environments even before the manufacture and use of the actual device. CFD techniques yield valuable insight into the VADs' effectiveness that enhances its operational behavior and decreases the risk associated with the use of the device, and at the same time decreases the manufacturing lead time and costs associated with the device. CFD has been quite beneficially used for performance evaluation of VADs, but it still needs further development as hemolysis and thrombosis models cannot be easily integrated with the flow simulations. This review article presents a brief introduction to rotary VADs with a primary focus on the current status of CFD analysis of axial VADs, which provide the most suitable methods for computational design and optimization of VADs. It also identifies critical knowledge gaps, outlines the hematological models and the difficulties encountered in integrating them into CFD. Finally, future work based on the current scenario is also included.

Keywords: LVAD; CVDs; Heart failure; CHF; CFD.

1. Introduction

A recent world health organization (WHO) report (1) implies that cardiovascular diseases (CVDs) are one of the major causes of mortality across the globe. Around 17.7 million people lost their lives because of CVDs in 2015 only, which is about 31% of global deaths. One of the significant health abnormalities that cause a substantial number of fatalities is coronary heart disease, which results in 7.4 million deaths annually (1). Heart transplantation, artificial organ substitution, mechanical circulatory assistance, or stem cell therapies are some of the pathways to treat a weak and/or poorly performing heart. Heart transplantation is a widely used treatment method to address the likely heart failure because of CVDs, but a shortage of donated hearts currently results in long waiting lists and high mortality (2,3). To overcome this issue, VADs in general and left ventricular assist devices (LVADs) in particular have become some of the preferable pathways for treatment to manage heart conditions (4,5).

The VADs are generally of two types, namely, pulsatile pumps and non-pulsatile or continuous pumps (Rotary). The smaller size, lower power consumption, less likelihood of formation of blood clots, and longer life are the main advantages associated with the use of a continuous pump over the pulsatile pump (6,7). Furthermore, the continuous pumps are of two types based on pumping mechanism, namely, centrifugal and axial pumps. Centrifugal pumps can produce high pressure rises at corresponding low flow rates, whereas axial flow pumps normally generate a higher flow rate with lower pressure rises (8). The axial blood pumps work at high rotating speeds to produce the required pressure rise and are more compact than the centrifugal pumps. The compact size and tubular design of axial flow pumps result in lesser implantation time, reduced cost of the process, and smaller invasion (9,10). The use of a continuous blood pump, which is a rotating machine with the heart which has a pulsatile nature of functioning, has given rise to a considerable amount of investigations to understand physiological and circulatory problems that may arise. Nose *et al.* (11) concluded that any circulatory or physiological abnormalities could be avoided if 20% more blood is flown through a continuous blood pump than that required in a pulsatile blood pump. However, the question of supremacy between the two is still a debatable topic among clinicians and researchers (6,12). Furthermore, continuous blood pumps are considered to be a better choice as they are more economical and reliable in many clinical conditions (11,13).

Heart patients have limited therapy choices. Due to a lack of heart donors, heart patients have trouble obtaining cardiac transplantation facilities (14). LVAD has emerged as a viable alternative for treating such types of heart failure. Significant efforts are undertaken to study the global and local flow characteristics of LVADs to optimize design and operation. CFD is a valuable tool for analyzing flow-handling engineering systems and improving biomechanical devices (15,16).

In this review paper, an overview of VADs and the current status of the axial LVADs designing through CFD is targeted. The existing literature on design, performance, hematological analysis, and optimization of commonly used axial heart pumps of non-pulsating nature is reviewed and analyzed. Further, the different design aspects and associated problems are also discussed. The future directions of the relevant parameters with consideration of hemolysis and thrombosis are also highlighted for LVADs application.

2. Heart failure

Chronic Heart Failure or Congestive Heart Failure (CHF) is characterized by the weakening of cardiac muscles and the resulting reduced contraction capacity (17). Insufficient blood supply due to CHF causes fatigue and shortness of breath. Insufficient blood flow pressure causes backflow to the heart, causing tissue congestion. This excess fluid can cause swelling in the legs, feet, and ankles, which is called "Edema." This fluid can also accumulate in the lungs, causing a pulmonary blockage (18).

Coronary artery disease may lead to CHF, diabetes, and hypertension (19). The US National Hospital Discharge Surveys show that this disease affects the elderly. This is the most common reason for hospitalization. More than 70% of people aged 65 or more hospitalized in America have CHF (20). The above indicates the extent of the problem that needs to be addressed through various mechanisms, of which VADs are an integral part.

2.1 Heart failure cases

CHF is now a global health issue. Globally, 37.7 million people have CHF, and the number is rising (21). WHO predicts that deaths may increase to 23.3 million because of cardiovascular diseases by 2030 (22). Initially, CHF was thought to be confined to the wealthy due to their sedentary lifestyle, which led to obesity, hypertension, and diabetes. However, the changing lifestyle of the growing middle class may soon expose them to CHF (23,24).

In the USA, approximately 870,000 new incidents and about 300,000 death cases because of CHF are reported annually (25). The survival rate of CHF patients is very low over a reasonably long duration (26). In the first year of the diagnosis, about half the number of diabetic patients having CHF die, and this number increases up to 70% in the 5th year (27). The above clearly indicates that interventions are needed to prolong the lives of CHF patients.

2.2 Available remedies

The treatment options for CHF patients vary from induced changes in the way of living, treatment, and trans-catheter interventions to heart transplants depending on the severity of CHF (28,29). Heart transplantation is the best treatment for CHF patients, but heart availability is limited. In the US, 2,100 to 2,500 donor hearts are available annually (25), while 3500 patients are still waiting for a heart transplant (28). The number of patients tends to increase each year, while donor heart availability is almost constant. Therefore, mechanical circulatory support devices can be used to bridge the gap between the need for hearts and those available from donors. As needed, these devices can be used for short or long periods of time (30). A device such as Total Artificial Heart (TAH) replaces the original heart and requires ventricle removal. However, a VAD helps a weak heart pump blood from the ventricle to the ascending aorta (Figure 1 and Figure 2) (31). The TAH is most appropriate for the patients in the final stages of congestive heart failure, for whom transplantation is not possible, whereas the VAD improves the pumping action of a failing heart (32,33) and hence can be used at different stages of the heart condition. Figure1 shows the CardioWest TAH, the only commercially available and FDA-approved total artificial heart for the bridge to transplant application (34). Its mechanisms include pneumatic actuation of the diaphragm, and it generates pulsating flow (35). Further, Dr. Richard Wampler and his team reported a novel TAH based on a pulsatile rotary pump. The main features of this TAH include no valves, no mechanical contact, single moving parts, lightweight, compact size, freedom for clinicians to adjust pressure and flow rate as per patient requirement, low hemolysis, and reduced thrombosis (36,37). Figure 2 shows a continuous VAD, Heartmate II. This is the world's most implanted VAD. Different components and flow conduits are also indicated (38). Both devices are reliable heart transplant alternatives. Because of its flexibility, a VAD has significant advantages over a TAH; for example, it can be used for long-term or short-term needs without removing the heart, and if the heart recovers, the VAD can be removed easily. The VADs are explored more in detail in the next section.

" Insert Supplementary figure here (Figure 1)" **and "[Insert Figure 2]"**

3. Ventricular Assist Devices (VADs)

The VADs are analogous to the mechanical pump, which supports the left or right failing ventricle without removing the native heart, and it may support, depending on the need, the right ventricle, left ventricle, or both together. The VADs may be used as bridge-to-recovery (BTR) in case of myocardial recovery, bridge to heart transplantation (BTT), or for the long-term use as destination therapy (DT) for patients having the requirement of a circulatory support system (42–44) and hence have wide-ranging application in managing a wide variety of heart conditions.

3.1 VADs classifications

VADs are generally classified by their outflow characteristics, either pulsatile/volume displacement or non-pulsatile/continuous. Other ways of classification are based on types of assistance needed, application period, therapeutic objective, and power supply (45,46). VADs are categorized by whether they support the left ventricle, right ventricle, or both, referred to as LVAD, RVAD, and BIVAD. The application period classifies VADs as short-term or long-term. Therapeutic objectives classify VADs as Series or By-Pass. The series type VADs take blood from the ascending aorta and deliver it to the descending aorta. These VADs operate in series with the heart and act in unison. Whereas By-Pass VADs pump blood from the left atrium or ventricle to ascending aorta, bypassing the mitral valve. The VADs can be powered by muscle or non-muscle power. Muscle-powered VADs use chest muscles to pump blood; non-muscle-powered VADs use an external power source. VADs are further classified by technological advancement as the first, second, and third generation. First-generation pulsatile VADs have many movable parts, a bulky structure, and low durability. Second-generation VADs are continuous and have rotating impellers to provide blood flow. The bearing of these VADs is submerged in blood or kept apart by seals. These VADs have leakages and sealing issues. Third-generation VADs use a magnetic bearing to support the impeller in the housing. This generation of VADs reduces the shortcomings of previous generations' VADs and is still being improved (42,43,47). In addition, researchers have reported a novel magnetically levitated device for VADs application called nutating blood pump or nutating disc pump based on the passive magnetic spherical bearing. Moreover, it showed acceptable levels of hemolysis and thrombosis for the predicted values of flow rate and wall shear stress (48,49). The broad classification framework for the VADs is shown in Figure 3. The VADs are mostly categorized on the basis of outflow characteristics which have been discussed in detail in the literature (44,50,51). The rotary blood pumps, being small in size, easy to implant, and have low rates of infections, are preferred in VADs (52). The following details of rotary VADs have been presented.

"[Insert Figure 3]"

3.2 Rotary type VADs

A rotary pump transfers energy to the fluid due to the impeller's rotation, and pressure is created at the pump's entry and exit. In these pumps, fluid enters the inlet through rotating impeller blades, which impart energy to the fluid in motion and produce continuous flow at the outlet. These pumps can precisely control the outflow by changing inlet and outlet conditions. The pumps are designed for maximum efficiency at design speed. The rotary pump may be categorized into axial flow pumps and centrifugal pumps.

The axial pumps require higher rotational speeds as compared to those of centrifugal pumps to generate the same pressure difference needed for the required flow rate (53). This high speed may increase shear stress and therefore increase the possibility of hemolysis. Nevertheless, higher rotational speed reduces the exposure time because of its comparatively lower volume, which is favourable. Though centrifugal pumps may be useful in low-speed applications, the significant advantages of the axial pump, like their smaller size and lesser power consumption, make these suitable for implantation in both adult and paediatric applications (9,54). This section has clearly highlighted that axial pumps have certain advantages over centrifugal pumps.

3.3 Characteristics of rotary VADs

Medical devices follow rigorous regulations for ensuring the safety of the patient, medical persons, and all other associated persons. The U.S. Food and Drug Administration (FDA) and the EC Medical Device Directive 93/42/EEC are the two most important regulatory bodies. The FDA is responsible for promoting a medical device on the basis of safety and effectiveness measures to the patients and also ensures the product labeling and instructions are followed by the manufacturers

(55,56). The different types of rotary VADs are summarized in Tables 1 and 2, shown below. The existing VADs have been classified on the basis of various indicators such as the location of the implant, flow rate, kind of assistance, duration, ambulation, and FDA approval status. Table 1 lists axial type VADs based on electrical actuation. These VADs are mostly implanted internally to support the left ventricle. However, Impella Right Percutaneous (RP) Systems are used to support the right ventricle for a short duration. Their flow rates within the pump range from 2.5 to 12 L/min. These devices can be implanted for long and short durations as per patient requirements. Moreover, these devices have very good ambulation. The VADs can also be categorized on the basis of flow regime, centrifugal, and radial type based on electrical actuation, as shown in Table 2. This category of VADs is mostly implanted internally for left ventricle support except for Medtronic Biopump. Medtronic Biopump can also be used to support the right ventricle, left ventricle, or both for a short duration application. The flow rate of such VADs also varies from 2.5 to 12 L/min. The patient's ambulation with such VADs except Medtronic Biopump was found to be satisfactory and can be used for long-term application.

Despite facing challenges in providing desirable functioning, the axial and centrifugal pumps are fast replacing the positive displacement pumps lately. VADs have great potential to support the weakened heart (57,58). None of these VADs treat heart failure holistically. These rotary VADs still face hemolysis, thrombosis, infections, and accurate flow control for pulsatility (52). Highly invasive techniques for implanting and removing VADs cause enormous surgical stress on patients. Efforts are being made to overcome the aforementioned limitations of VADs through miniaturization and less invasive techniques. As a result, there is a greater need for advancements in existing VAD technology. Making use of the advanced computational techniques, the performance of the existing VADs can be enhanced by improving design parameters and flow characteristics.

Table 1: Axial flow types (VADs) based on electrical actuation (59,60).

S No.	Approved VADs	Location	Flow Rate (L/min)	Assistance	Duration of use	Ambulation	FDA STATUS
1.	Jarvik Heart Jarvik 2000 Flow Maker	Internal	7	LVAD	Long-years	Good	In trial
2.	Micromed DeBakey	Internal	10	LVAD	Long-years	Good	Not approved (in the trial)
3.	HeartMate II Left Ventricular Assist System	Internal	10	LVAD	Long-years	Good	BTT/DT
4.	Berlin Heart Incor	Internal	7	LVAD	Long-years	Good	Not approved (in the trial)
5.	Impella Right Percutaneous (RP)System: H140001	Internal	> 4	RVAD	14 days	Good	Short term use

6.	Impella 2.5 System - P140003	Internal	-	Temporary ventricular support device	>6 Hours	Good	Short term use
7.	Impella RP® System - P170011	Internal	≥4	RVAD	14 days	Good	Short term use
8.	Impella Ventricular Support Systems - P140003/S018	Internal	2.5-5	LVAD	06 days	Good	Short term use

Table 2: Centrifugal and radial type based on electrical actuation (VADs) (59,60).

S No.	Approved VADs	Location	VADs Types based on working	Flow Rate (L/min)	Assistance	Duration of Use	Ambulation	FDA STATUS
1.	Medtronic Biopump(80 ml &48ml)	Extracorporeal	Radial flow	10	LVAD/RVAD/ BIVAD	Short for <5 days	No	BTR/BTT
2.	Arrow Int.CorAide	Internal	Centrifugal (radial flow)	3.3 to 4.9	LVAD	Long-years	Good, discharge possible	Not approved (in the trial)
3.	Impella Recover LP2,5 & LP 5,0	Internal	Radial flow	2.5 and 5	LVAD	Short for 5-7 days	Fair-short transportation	Not approved (in the trial)
4.	Terumo DuraHeart	Internal	Centrifugal	--	LVAD	Long-up to 5 years	Good	Not approved (in the trial)
5.	Ventracor VentrAssist	Internal	Centrifugal (radial flow)	12	LVAD	Long-years	Good	Approve for the pilot study

6.	World Heart HeartQuest VAD	Internal	Centrifugal (radial flow)	--	LVAD	Long-years	Good	Preclinical
7.	HeartMate 3™ Left Ventricular Assist System - P160054	Internal	Centrifugal	10	LVAD	6 Months	Good	BTT/DT
8.	HeartWare™ HVAD™ - P100047/S090	Internal	Centrifugal	10	LVAD	1-2 Years	Good	BTT/DT

4. Computational techniques

CFD is an excellent tool in developing VADs, allowing different designs to characterize and optimize in-silico performance. A CFD analysis of a computational domain includes preprocessing, solver, and post-processing. In preprocessing, the problem's geometry is created. Depending on the problem, geometry may be designed using equations or computer-aided design (CAD) software. Generally, Complex geometry is created by CAD software. CAD geometry can be imported into CFD analysis software with a suitable file format. Meshing or grid generation discretizes the geometry in small elements. The physics of the problem and the material properties to be analyzed are specified after meshing. Material properties like viscosity, density, thermal conductivity, etc., of working fluids must be carefully specified as they affect the results. Boundary conditions (BCs) are specified by the physics of the problem. A stationary wall in the flow domain is defined as a no-slip wall condition. Moving wall BC is applied to walls in translation or rotation. The initial BC for the Inlet is usually a velocity, pressure, or mass flow rate. When the pressure at the outlet is known, the outlet is generally defined as a pressure outlet; otherwise, outflow BC is used. The solver part of the CFD simulation consists of solving the equations which govern the physics involved in the problem iteratively. The CFD simulation process is depicted as a flow chart shown in Figure 4. For this, different algorithms are used to solve the partial differential equations; one such algorithm is SIMPLE (61). Thereafter, the results are analyzed in the post-processing part of the CFD analysis.

“[Insert Figure 4]”

Commercial CFD packages solve the governing equations, continuity, and momentum equations of the flow field by finite volume (62) or finite element approach. User-defined functions

(UDFs) allow compiling inlet profiles and wall functions like log-law function, velocity or mass flow rate, etc., which are available at the boundary condition options. These UDFs help in carrying out the simulations with variable conditions at any boundary of the domain. Available literature suggests most VADs have been analyzed using commercial CFD codes with varied success.

ANSYS, Inc. (Canonsburg, PA) and STAR-CD of CD-Adapco (Computational Dynamics-Analysis and Design Application Company Limited) are some of the popular commercial CFD codes available for the researchers (63,64). ANSYS comprises Fluent (65–68) and CFX (69–77) solver options for analyzing fluid flow characteristics of VADs or flow problems of interest. Many other CFD software packages are used by some researchers, like product AcuSolve of ACUSIM Software (64), Adina R&D (Watertown, MA), etc. Specific CFD solvers have also been developed by some researchers for solving the specific flow problem of VADs. An alternative approach for flow analysis of the GYRO pump, where blood is considered as a viscoelastic fluid to capture the shear-thinning properties based on Deformable-Spatial-Domain, was used by Behr *et al.* (78,79). The pressure head calculations were found to be in line with that of experimental results showing an error of 5% normally, which increased to 12% at high speeds.

Various studies have been carried out on VADs for flow analysis with CFD software using a wide variety of approaches. For the analysis of VADs, the methods such as Smoothed particle hydrodynamics (80), Spectral methods (81), and the Lattice Boltzmann method (82,83) have also been explored. These methods may find use in the in-depth analysis of the new class of VADs in the future (84,85).

CFD equations are solved iteratively. Convergence is important for residual-based findings. Different researchers assess convergence criteria differently, but the most typical

technique is to monitor residuals, such as when they drop below a threshold or become static. The relative importance of different flow variables affects convergence criteria. These computational algorithms utilize specific grid generation, problem-solving, and system approximation methodologies. So, proper computational methods are crucial for the CFD analysis of VADs. The CFD analysis begins with modeling the geometry and flow domain which is explored in the following section.

4.1 Geometry representation

The compactness of axial blood pumps makes them popular for adult as well as pediatric heart patients. The axial blood pumps have three main parts, namely a straightener, impeller, and diffuser, as shown in Figure 5. The use of a straightener helps to avoid the pre-rotation of the fluid before entry to the impeller. The impeller rotates and transfers the energy to the blood, and this energy is converted into the pressure head by the diffuser (86). However, some axial pumps do not have a diffuser for the simplicity of design and manufacturing (66,87). The axial pumps may be of two types on the basis of diffuser configuration. In the first type of axial pump, diffuser blades are connected with the pump hub just as in HeartMate II and DeBakey VAD (88,89), and in other types, the diffuser blades are not connected with the pump hub like Magnevad and LEV-VAD (71,90)**.** The Impeller and diffuser play a significant role in deciding the performance characteristics of an axial blood pump (77,86,91–93). Current VAD geometries are generally created in CAD software because of design complexities. The VADs should be imported to the CFD environment in a compatible file format to analyze if the geometry is created in CAD software. Generally, the geometry consists of solid parts of the pump, out of which the portion where blood flows is extracted through boolean subtraction to arrive at the flow domain. Precise modeling is very important for predicting the performance characteristics and other relevant parameters for implanting the VADs. The other most critical issues are to find optimal outflow cannula positions and configuration during pump implantation. The 3D patient-specific model of the aorta and pump geometry will require to address these issues through CFD simulations. The 3D model is extracted from computed tomography-scan slices through medical imaging techniques (94). The consideration of these geometrical features and their parameters influence the flow characteristics inside the VADs. Thus, the effects of geometries may play a significant role in the advancement of VADs technology.

"[Insert Figure 5]"

4.2 Axial blood pumps

Axial blood pumps have been explored as a viable option to manage the condition of heart patients living with severe heart ailments. It is also established as an effective treatment option used for assisting heart failure patients with different severity levels of the ailment (95–97). Usually, low shear rates result in platelet adhesion and thrombus, whereas high shear rate values result in platelet activation as well as hemolysis (98–100).

The impeller radii of axial flow blood pumps vary from 2-10 mm, which may produce flow rates of 1.5-6 L/min and a pressure head range of 50-140 mmHg at the operating speed of 6000-45,000 rpm (28). In the following, some important pumps and their application has been described.

The first axial flow pump to be developed was known as a Hemopump which comprised of an inlet cannula, a screw rotor, and a flexible drive cable. The screw rotor is mounted at the end of the drive cable in the cannula, and the drive cable is rotated by an external motor. The friction between the drive cable and bearing is reduced by using lubricating fluid. The rotor is rotated by the drive cable, which is driven by an external motor (101). The Hemopump was operated in 1988 for the first time (102), and Impella was used in 1999 for clinical purposes, which was similar to Hemopump in design and operation (103). The Hemopump and Impella have been primarily used for temporary support, which needed lesser surgical procedures (104,105). MicroMed DeBakey was the first pump to be used as long-term axial VAD as the bridge to transplant and was employed in 1998 for clinical purposes. The modern version of this VAD is HeartAssist5. The Jarvik Heart® (106,107) was implanted in 1999 for the first time as destination therapy (108), whereas HeartMate II (9,109) was implanted in 2000 for the first time (110). The components of these pumps included pivot bearings, fixed axels together with sealed, brushless electromagnetic motors (111,112). These pumps reduced the contact surface and hence alleviated the lubrication requirement. The flow tube of these pumps included a fixed front straightener, an impeller only as a moving part, and a fixed diffuser. The bearings of these pumps are made up of ceramic components which evolve low wear and long durability. However, the blood immerged bearing used in these pumps stimulates thrombus formation (113). The pump speed can be regulated in Jarvik Heart® and HeartMate II, whereas the performance of HeartAssist5 can be managed and checked remotely (114). The flow of these pumps is evaluated using linear relation of current-flow over its all working range. However, the current-flow relation is not linear overall working range of the axial pump. This aspect limits the accuracy of flow calculation, mainly at low rates (115). For instance, the flow rate below 3.0 L/min does not appear in the HeartMate II. The additional sensor may be used in the outlet conduit of the pump to measure the flow rate. However, these additional components may involve complexity enhancement, cost increment biocompatibility, and sensor failure. For instance, HeartAssist5 estimates the flow rate by using an ultrasonic probe at its outlet conduit. Therefore, various flow estimation models were suggested by the different researchers based on already known parameters of the pump controller (116). However, these models face challenges in predicting accurate flow estimation. Further, flow pulsatility is desired in patients under continuous pump support to prevent undesired outcomes, which can be achieved by an accurate speed modulation algorithm (117). However, the accuracy of flow estimation is needed for the accurate performance of the flow algorithm for achieving pulsatility (9).

For pediatric support, Jarvik started trials on pediatric VAD in 2010 and was awarded the license by FDA in 2013 (Closing in on the Pump KIN Trial of the Jarvik 2015 ventricular assist device) for supporting such works (118). For avoiding most of the contact surfaces in this pump, hydrodynamic bearings are being used. In this pump, the spindle of the pump having two impeller blades are assisted through an inflow and out-flow tripod when the impeller is stationary. As the impeller starts rotating, lift forces are initiated between the impeller and the tripod tips that are sufficient to assist the impeller (119). The magnetically levitated axial flow pump, named INCOR pump, was first used in 2002 for heart patients. The energy consumed by the pump is 3–4 W, whereas 0.6W energy is used to maintain magnetic suspension (120). This pump gives a specific advantage as it uses a noncontact impeller suspended by a magnetic bearing, while other available VADs use a blood-emerged bearing (121). The magnetic system increases the weight and size of the pump more than other bearings systems, but at the same time, it provides better hemocompatibility for long-term use (122). This aspect may be optimized for further miniaturization.

In this section, the tubular size (length x diameter) and other performance-related issues are discussed. The Jarvik Heart® is comparatively small in size (78mm x 26mm) than other commercially available continuous pumps, whereas the INCOR pump has the longest length

(120mm x 30mm) than other pumps. The other pumps, HeartMate II and the HeartAssist5 have their size (81mm x 43mm) and (71mm x 30mm), respectively (123). The miniaturization of VAD is desirable to achieve thoracic implantation in all categories of patients (124). These pumps, except the INCOR pump, use contact bearing, which raises friction, heat generation, pump failure, and thrombosis. Moreover, these axial pumps produce high-pressure differences across the pump, which may cause suction events and ventricle collapse in case of abruptly reduced filling pressure. The ventricular preload and position of the inflow cannula play an important role in addressing these issues (52). For instance, a new implantation technique was suggested by the researcher for the inflow cannula of the Berlin Heart INCOR system, which reduces the time of implantation and thrombosis (125). However, these pumps provide improved life of the patient with mobility and help in myocardial recovery compared to previous generation pumps. Furthermore, the other limitations are their driveline infection, use of extracorporeal power source, hemolysis, and demand for anticoagulation therapy as they are prone to thromboembolic events (126,127). Currently used pumps vary in their design based on the impeller, bearing, and methods of actuation (126). The geometric configuration of the impeller affects hydraulic efficiency as well as blood stagnation regions (128), and the structure of pumps highly influences the blood flow characteristics inside the pump (129,130).

Thus, the overall performance of the pump depends on the selection of an appropriate geometry, type of bearing used, flow regime, flow path, and blood-contacting surfaces. These parameters are required to be taken into consideration for further enhancement using the CFD tool, which may bridge the problems associated with current pumps.

4.3 Meshing scheme

Mesh independent outcomes are necessary for understanding the performance characteristics of VADs. Meshing parameters depend on the pump geometry, and different software have different types of meshing tools and thus may influence the results of the pump under consideration. Different types of mesh elements are used in CFD analysis. CFD analysis of the computational domain must include grid independence tests to eliminate the error because of mesh parameters and discretization (131–133). But complete grid-independent tests are reported rarely, and results have been shown with very little information about mesh densities and convergence criteria (134,135).

Most of the CFD software packages have the capability of automatic meshing and generating tetrahedral or hybrid mesh. A smooth and effective simulation of the flow problem in the solver needs a lot of manipulations during mesh generation. Some of the meshing software is dedicated to the meshing of turbomachines, which have the ability to generate meshes (O-grid) near the blade region of the blood pumps for computing flow variables (136,137). Generally, an H-grid mesh is generated between two blades. The mesh size is generally kept very small near the blade because of high gradients, while it is generally kept reasonably coarse between the blades to manage memory requirements during the computational process. The hexahedral grids have also been used, for example, these were generated by two different methodologies for PVAD3 for the impeller region, and a comparison was made to show which type of mesh should be chosen as per geometry suitability (72). The better mesh quality was obtained by focusing on the regions between the blades rather than around the blade's profile. Wu *et al*. (138,139) have developed a program for rotary pumps and blood shearing instruments for generating mesh based on the elliptic method. It creates hexahedral grids of high quality and ensures boundary orthogonality. The meshing software generally uploads geometry from CAD software as surfaces, and Wu *et al.* (138,139) used parameterized geometry creation method by the use of curves, which generated auto-enabled mesh and was capable of updating it as per geometry modifications. Also, recently a new technique has been developed by researchers for automatic hex dominant mesh generation for CFD application (140). The skewness of mesh was high at abruptly changed regions of the geometry, but overall good mesh quality was obtained even for complex geometry. The Shear-slip mesh update method (SSMUM) is an alternative method for such devices. Qu et al. (141) suggested improved SSMUM, which is instrumental for rotating machines, by providing conservative, smooth and high-accuracy flow fields.

The further enhancement on automatic hex dominant mesh and improved SSMUM are yet to be explored for computational studies of axial blood pumps. These mesh methods may be used to enhance the accuracy of the flow fields and solution. Further, the comparison of computational time taken by these methods is crucial in choosing the most suitable one.

4.4 Physical setup and boundary conditions

Accurate simulation of the flow process in CFD requires specified parameters related to the physics of the problems, such as the flow environments and fluid properties. These components are shown in Figure 4; the flowchart highlights the various flow physics for the simulation setup. CFD process starts with the created or imported geometry in the best possible way, and the then concerned domain is separated into the smaller part for grid or mesh generation. The solver and solution parts are followed to find the different targeted results. The solver part of CFD simulation includes several physical options as conditions of the wall, frame of reference, fluid modeling, and other boundary conditions. Boundary conditions are used to define flow conditions at entry and exit as well as outer wall and inner obstruction (142). Blood is considered a non-Newtonian fluid that exhibits the behavior of shear-thinning (143). Blood follows Newtonian nature with fluid viscosity of about 3.4-4 mPs if the shear rate becomes greater than 100 s^{-1} (144,145). Velocity or pressure is the usual inlet, and pressure or outflow is the usual outlet boundary conditions of the heart pump to simulate the flow characteristics of mechanical circulatory support. It has been observed that steady flow conditions are assumed, notwithstanding the unsteady pulsatile nature of blood flow at the inlet. CFD studies on axial VADs are used for evaluating hydraulic performances, thrombosis, and hemolysis predictions.

The boundary conditions and different turbulence models used by various researchers for axial flow blood pump simulation are summarized in Table 3. Generally, CFD simulations are carried out by setting inlet and outlet boundary conditions for a constant speed of an impeller with an appropriately selected turbulence model. Most of the researchers analyzed the complete part of LVAD by considering steady-state simulation with blood taken as Newtonian fluid and applied the k-ε turbulence model for the analysis, as explored in the following Table 3. The unsteady or transient CFD simulations for the analysis of LVAD with consideration of blood as Newtonian fluid (67,132,146–148) or as non-Newtonian fluid were also analyzed by researchers (149,150). These considerations deviate from the accurate simulation of the blood pumps. A few CFD simulations were carried out with consideration of blood flow as two-phase (Red blood cells and Plasma) flow using a multiphase model (135). The rotating impeller is simulated in CFD by two approaches, multiple reference frames (MRF) and sliding mesh, for calculating flow through the blood pump. The MRF approach approximates the impeller motion by specifying the two regions which move with a constant velocity relative to each other. It is based on steady-state assumptions, and this approach cannot capture unsteady effects. In the most recent CFD analysis, the rotating impeller has been assigned MRF or frozen rotor option to capture the flow visualization. The second approach, a sliding mesh, is a more realistic approach in which the relative velocity of two fluid zones alters with the position of the mesh at every time step. But the sliding mesh approach increases the computational cost. The CFD simulations in which the rotating impeller was assigned as moving mesh or sliding mesh were also analyzed by a few researchers (38,67,131,146,148). Recently, Wang et al. (151) analyzed the influence of mixing plane, frozen rotor, and sliding mesh methods on simulation by comparing the simulation results with PIV data. The results showed that frozen rotor and sliding mesh methods gave more reliable results than the mixing plane method. So, the use of these two methods is recommended during the simulation of the axial blood pump. The stationary walls or outer case were assigned as no-slip boundary conditions during the simulation. However, some of the analyses were carried out by creating a closed mock loop for a constant speed of the impeller, and the same was validated with experimental data. The blade profile of the impeller and diffuser were analyzed by various investigators for enhancement of the overall performance of the LVAD. Nammakie et al. (6) used variable speed (speed modulation) of impeller for generating pulsatile flow by a continuous LVAD, though this aspect was analyzed by only few researchers. Most of the researchers analyzed only the conventional design of axial blood pumps, whereas the hubless pump design is not explored well, which may give the new possibility for future device development (152).

The turbulence models used by the researchers have their limitations in terms of application and accuracy. The incorporation of the multi-component characteristics of blood is emphasized for better understanding and evaluation of the thrombosis and hemolysis phenomenon. However, the consideration of blood cells (in millions) and their properties is a great hurdle for computational hardware. Therefore, researchers frequently use the Newtonian and single-phase fluid in CFD analysis of blood pumps. Further, the numerically calculated hemodynamic parameters are rarely validated by the researchers because of the complexities of measuring them experimentally. The researchers use different boundary settings, grid generation methods, and solution procedures which directly affect the accuracy of results. These aspects increase the uncertainty of choosing the right selection criteria for the analysis of blood pumps. A comparative study may help to select a particular setting during CFD simulation for more accurate results. Furthermore, the profiles of blades of the pump impeller, diffuser, and profile of rotational speed needed for generating pulsatile flow can be optimized in terms of hydraulic and clinical performance in the near future.

Table 3: Boundary conditions and turbulence models used by various researchers for CFD studies of Axial-VADs.

Author(s)	**Part Analyzed**	**Flow Boundary Condition**	**Rotation of impeller, N (rpm)**	**Turbulence Model**	**Conclusion**
Apel *et al.* (153)	Whole LVAD	Inlet = Constant velocity, Outlet = Static pressure	**N** =25000,30000 and 32500	k-ε	A computational model of the Microaxial blood pump was analyzed for quantification of hemolysis using the CFD tool. It was suggested that proper turbulence models need to be selected for the hemolysis prediction. However, these predictions were not compared experimentally.
Mitoh *et al.* (154)	Impeller with guide vane	Inlet =50 mmHg Outlet=5 L/min	**N** =7000	k-ε	Hemolysis in the axial flow pump was predicted by CFD analysis. The results were shown a good correlation with measured values.

Yano *et al.* (155)	Whole LVAD	Inlet =50 mmHg Outlet=5 L/min	**N** =7000-8000	k-ε	Hemolysis estimation, by CFD analysis and measured using bovine blood shown a good correlation.
Chan *et al.* (156)	Whole LVAD	Inlet =Not specified Outlet= Not specified	**N** =10000	-	The hydraulic performance of the axial flow pump was presented. Two inducer-impellers and three diffusers were analyzed in a closed-loop circuit. Based on the results, it was seen double start inducer-impeller showed good results. Stagnation and reversal zone was shown in CFD results. Modifications in design and blade profile may eliminate these problems.
Untaroiu *et al*. (71)	Whole LVAD	Inlet = Constant velocity (Q = 2-10 L/min)	**N** =5000-8000	k-ε	An axial-flow ventricular assist device with a magnetically levitated impeller (LEV-VAD) was designed. Flow testing was performed on the

		Outlet = 20000 Pa			plastic prototype of LEV-VAD, and it was compared with CFD predictions.
Untaroiu *et al.* (157)	Whole LVAD	Inlet = Constant velocity (Q = 2-10 L/min) Outlet = Static pressure	**N** =5000-8000	k-ε	The CFD analysis of magnetically levitated impeller (LEV-VAD) with six and three diffuser blades was done to improve the performance of the LEV-VAD.
Throckmorton *et al.* (158)	Whole LVAD	Inlet = Constant velocity (Q = 0.5-3 L/min) Outlet = 20000 Pa	**N** =7000-9000	k-ε	A new axial flow pediatric ventricular assist device (PVAD) was designed. This was the optimized design of the previous PVAD1.CFD analysis and experimental testing were used for the optimization of the pump.

Triep *et al.* (159)	Impeller of LVAD	Mock loop	**N** =45000-50000	k-ε	Three impellers were designed, and performance was examined by CFD analysis and digital particle image velocimetry. Both the results have shown good conformity.
Throckmorton *et al*. (72)	Whole LVAD	Inlet = Constant velocity (Q = 0.5-4 L/min) Outlet = 20000 Pa	**N** =6000-9000	k-ε	Three pediatric ventricular assist devices (PVAD), PVAD1, PVAD2, and PVAD3, were examined by experimental analysis and by CFD analysis. The PVAD3 created by design adjustment in PVAD2 showed good performance to fulfill the design objective.
Chua *et al*. (67)	Whole LVAD	Inlet : 0 mmHg Outlet:80,100,120 mmHg	**N** =10000,11000 and 12000	k-ε	Flow characteristics of axial blood pumps are computed on commercial Fluent software. The pump was able to produce 5.14 L/min flow rate at 100mmHg when operating at 11000 rpm.

Yao-jun and Fu-jun(160)	Whole LVAD	Inlet: Constant velocity Outlet: Velocity Gradient component =0	**N** =1450	k-ε	CFD analysis of the Axial-flow pump with an inducer was examined. Simulations results showed that maximum pressure was generated at 0° or 30° of alignment of the inducer with the impeller of the pump. These results have shown good agreement with the experimental data.
Throckmorton and Untaroiu (73)	Whole LVAD	Inlet: Constant velocity (Q = 0.5-5 L/min) Outlet: P= 20,000 Pa(150	**N** =6000-8000	k-ε	An axial flow pediatric ventricular assist device (PVAD4) was designed and developed. Its performance-improvement was compared with the previous design (PVAD3) by CFD simulation.

		mmHg)			
Zhang *et al.* (161)	Whole LVAD	Circulating loop	**N** =8000-10000	k-ε	CFD analysis and in vivo testing had examined for the newly developed axial flow VAD in Beijing. The in vivo and in vitro performance was enough for an LVAD. The hemolysis was evaluated in the in-vivo test slightly more but within tolerable limits.
Hui-min *et al*. (133)	Whole LVAD	Circulating loop	**N** =8000,9000	k-ω	The developed axial flow pump is shown high efficiency with a lightweight and compact design. The CFD results show good agreement with experimental results.
Hui-min *et al*. (162)	Whole LVAD	Circulating loop	**N** =8000,9000	k-ω	The design of the outlet guiding vane, bend angle selection at outlet connection, and flow channel is improved by numerical simulations. These results were verified by Particle image velocimetry (PIV) tests.

Su Boyang *et al.* (163)	Whole LVAD	Inlet: Velocity Outlet: Pressure	**N** =9500	(SST) k- ω	The axial flow blood pump is evaluated for both CFD and PIV results. These results were compared for validation.
Wu *et al.* (4)	Impeller	Inlet: Constant Velocity (Q=0.1055kg /s,) Outlet: =110 mmHg	**N** =7000	k-ε	Two types of impeller blades, one with constant thickness blade and other variable thickness blades, were simulated. The simulation results have shown that the variable thickness blade is shown better results than a constant thickness blade impeller.
Wu *et al.* (164)	Whole LVAD	Inlet: Constant Velocity (Q = 0.3-2.3 L/min) Outlet: 100	**N** =10000-16000	SST	CFD-based design optimization was used for pediatric VAD. The optimized VAD, PF4, had a novel impeller with the first stage of mixed flow; after that, a short axial flow was the second stage. Experimental and CFD results showed excellent hemodynamics performance with low hemolysis.

		mmHg (pressure head given)			
Qi *et al.* (165)	Impeller	Static pressure difference at inlet and outlet: 80-100mmHg	**N** =20000-	k-ε	CFD analysis of axial blood pumps with three different heights of the impeller. The outlet flow and scalar shear stress increased with an increase in the height of the impeller. The Impeller with a height of 2 mm was shown acceptable shear stress with a flow rate of 5 L/min at a speed rotating 20000 rpm, while the pressure difference 100mmHg.
Carswell *et al.* (149)	Whole LVAD	Inlet: Pressure(0 Pa) Outlet: Pressure(330	**N** =2000,25000	k-ε, k-ω	Hemolysis prediction of micro axial blood pump (μLVAD) computed by CFD simulation and predicted results were compared with experimental data of other literature.

		0-2000000pa)			
Tie-yan *et al.* (135)	Whole LVAD	Circulating loop	**N** =8000	(SST) k- ω	CFD multiphase (plasma free hemoglobin content and the hematocrit) simulation of axial flow impeller blood pump NIVADIII was carried out, and hydrodynamics performance was obtained to calculate hemolysis. These results were compared with the experimental calculation of hemolysis.
Su *et al*. (86)	Whole LVAD	-	**N** =9000-10000	(SST) k- ω	Three different diffusers were analyzed by CFD simulation. Each of the diffusers was analyzed for flow rate 4-7 L/min and rotational speed 9000-10000 rpm. These computational results were not validated or compared with any experimental data.
Thamsen *et al.* (166)	Whole LVAD	Inlet: Constant	**N** =6000-15000	k-ω	Numerical analysis of HeartMate II and HeartWare HVAD rotary blood pumps were

		Velocity (Q = 4.5 L/min) Outlet: 80 mmHg			examined to compare flow fields and hemocompatibility of the pumps. This study may be able to help a new generation of blood pumps with reduced blood trauma.
Schule *et al.* (148)	Whole LVAD	Inlet: Constant Velocity (Q = 4.5 L/min) Outlet: Zero gradient	**N** =7000-9000	k-ω	CFD analysis and experimental analysis of the HeartMate II pump were performed with respect to hemolysis and thrombus formation. The recirculation and flow separation zones were found on the impeller blade leading age and diffuser. These are the critical place for thrombus formation.
Nammakie *et al.* (6)	Whole LVAD	Inlet: Pressure= 40mmHg Outlet: Aortic	Variable rotational speed (3700-7500)	k-ε	The computational results of axial flow VAD, shown the sinusoidal rotational speed pattern of the impeller, are more favorable than the trapezoidal one. The effect of these patterns on

		pressure pulse of the patient.			the geometric configuration of the VAD may be studied further.
Selishchev *et al*. (167)	Whole LVAD	Inlet: Constant flow rate Outlet: Static pressure Circulating loop also used.	**N** =5000-9000	k-ω SST	The sputnik design axial pump was developed by optimizing the sputnik one design. The pump weight, length, and diameter reduced were reduced, which reduced the power consumption of the pump by 15%. The CFD analysis was performed for the operating condition as well as for the physiological conditions also without considering hemolysis and thrombosis.
Liu *et al.* (168)	Whole LVAD	Inlet: Static pressure=10 mmHg Outlet: Constant	**N** =9000-12000	k-ε	An axial left ventricular assist device, LAP31 for partial assist heart failure patients, was developed with the novel structure of cantilevered main and splitter blades of a diffuser. This blood pump was designed to deliver a 3 L/min flow rate with

		flow rate (Q = 1-8 L/min)			pressure rises of 100 mmHg. CFD analysis of LAP31 was performed to evaluate hydraulic and hemodynamics capabilities. In vitro test was also conducted.
Sang and Zhou (146)	Whole LVAD	Inlet: Velocity Outlet: Outflow	**N** =7000	k-ε, RNG k-ε k-ω, and k-ω SST	Hydraulic performances and pressure fluctuations of two-stage axial blood pumps were evaluated numerically and experimentally by varying guide vane outlet angles. At the outlet angle, 10° efficiencies and pressure head were optimum.
Liu *et al.* (147)	Whole LVAD	Inlet : Static pressure=10 mmHg Outlet: Constant flow rate	**N** =9000, 10000	k-ε	An axial blood pump, LAP30, with variation in blade radial gap, was modeled and simulated numerically for the hydraulic performance evaluation. These results were also tested experimentally. The efficiency was decreasing, whereas the pressure head increased with an increase in gap 0.1 to 0.3mm, and The

		(Q = 1-5 L/min)			quantification of hemolysis was also performed numerically.
Kannojiya *et al.*(150)	Whole LVAD	Inlet: Static pressure Outlet: Static pressure	**N** =6000-10500	k-ε, RNG k-ε k-ω, and k-ω SST	An axial blood pump model proposed by researchers which shown improved hemodynamic performance over the previous model investigated computationally. The Bird-Carreau model considered for blood flow and the transient blade row model for pump hemodynamics were adopted to make more realistic simulations. The experiment on its physical design may be conducted for the proper validation of computational results.

4.5 Turbulence

The flow of blood through the pump is mainly turbulent due to the high rotating speed of the impeller and curvature effects, consisting of vortices that carry the energy gained via the main flow to the Kolmogorov scale and convert it into heat. The effects of turbulence should be considered for the correct estimation of the loss in pressure and the shear stress estimation, which has a direct relation to the prediction of thrombosis and various other blood cell damages. This makes the choice of the turbulence model for the blood pump simulation very critical. The flow of blood through the VADs is generally characterized by a low Reynolds number (Re) depending on its specifications. The difficulty in selecting the turbulence model arises due to the fact that most of the turbulence models are developed by considering high Re number flows. The Re number of axial blood pumps at the inlet varies from 2200-5500, whereas for the impeller region, it ranges from 43000-51000. However, the Re number of Impella pumps at the impeller region ranges from 13000-20000. The Re numbers of the impeller region are greatly below the threshold value 10^6 (28), which implies the investigator chose the laminar model. However, the diffuser region of pumps is associated with flow separation, high vortices, and turbulence generation, for which a turbulence model would give better results than a laminar solution (76). The small size and complex geometry of the blood pump make the wall effects more crucial, which greatly influence the flow fields. The feasibility and accuracy of CFD simulation depend on near-wall treatment methods, limitation of grid arrangement, and their agreement with selected turbulent models. The turbulence models provide good results when a defined turbulent boundary layer is discretized near the wall, and a fully developed flow exists away from the wall. On the other hand, laminar simulation, with a dense mesh in the boundary layer, is preferred for solving the flow which are dominated by viscous forces.

The k–ε turbulence model has been extensively used for blood pump design (76,154). The k and ε show the turbulent kinetic energy and turbulent dissipation rate, respectively. Considering the eddy viscosity assumption to relate the Reynolds stress and turbulent terms to the mean flow variables, the k–ε turbulence model have a few limitations, i.e., poor accuracy for low Reynolds numbers or inefficiency in capturing the fluid characteristics during flow separation along the boundary. k-ε model is used extensively in CFD study of the centrifugal blood pumps (169–171) as well as for axial blood pumps (4,6,67,71–73,146,147,149,150,153–155,157–161,165,168) as shown in Table 3. The k-ε model is a two-equation model to solve the transport of the kinetic energy of the eddies and the dissipation rate of the eddies along with the Navier Stokes and continuity equations. k-ε model is based on empirical relations for the calculation of ε, but the derivation of the k is based on an exact equation. The k-ε model finds suitability in industrial applications for its robustness, acceptable accuracy, and computational efforts for a range of flows (150). The flow is assumed to be fully turbulent, and the negligible molecular viscosity effects render it more suitable for high turbulent flows only (172). There are several other limitations also in the k-ε model, such as the growth rate of the turbulent kinetic energy is over predicted, reattachment length in the wake of backward-facing step in turbulent flows, and rotating channel flows the k-ε model incorrectly predicts an asymmetric mean velocity profile as symmetric one (74,173).

Another important two-equation RANS model is known as the k-ω model. Here ω stands for the turbulence frequency. This model gives better output for wall-bounded, and low Reynolds number flows. The k–ω model switches automatically from a wall function to a low Reynolds number flow according to the mesh spacing (174). To capture the separation phenomenon in a boundary flow, the k-ω model meets the expectations better than the k-ε model. SST k-ω model is a combination of the original k-ω model and standard k-ε model, switching to k-ε for the fully turbulent region and back to k-ω for near-wall and low Reynolds number flows (62,175). Al-

Azawy *et al.* (176) compared six turbulence models for a pulsatile heart pump numerically, and the results were compared with experimental data. The RSM and SST k-ω model showed close agreement with reference data. The large eddy simulation (LES) turbulence model is also being used for numerical analysis of different rotating machines, which gives acceptable results. However, the LES model needs high computational cost than commonly used models (177). Recently, Wang et al.(178) computed six turbulence models for an axial blood pump numerically, and the error of velocity fields was compared with PIV data. The SST k-ω model showed the smallest error and close agreement with experimental data. However, the LES model was not considered in this study.

The discussed turbulence models are capable of predicting the pressure difference well, but they show huge variations in the prediction of velocity fields, which play a very significant role in blood pump design configuration and blood damage analysis. The SST k-ω model is advised for computational analysis of axial blood pumps. Furthermore, the LES model or advanced turbulence models may be examined and compared with commonly used turbulence models, which may give more precise results than these models.

5. Hydraulic performance parameters of VADs

Hydraulic performance evaluations are the prime use of numerical simulations in the development of VADs system and their components. Generally, the calculation starts with the pressure head; after that, hydraulic power, hydraulic efficiency, hemolysis, and thrombosis performance are evaluated. The main function of the blood pump is to supply blood at the appropriate pressure and flow rate. Thus hydraulic performance is quantified through different terms, namely pressure ratio, flow rate, and hydraulic efficiency. The numerical simulation process includes many assumptions, approximations, and simplifications during the analysis of blood pumps. Therefore, these quantities need to be established through experiments and/or CFD simulations (28,86,147). In this connection, validation and verification of CFD findings become necessary. Further, these parameters are evaluated to test whether the blood pumps meet the physiological needs of heart patients and also help to compare the performance of different types of VADs. The required parameters are discussed in detail underneath.

5.1 Pressure head

The work performed by the pump on the working fluid is measured by the difference of the energy (pressure if the flow velocities are the same at inlet and exit) at entry and exit of the pump, named as pressure head. The H-Q curve, which is the plot between the pressure head and corresponding volumetric flow, shows the performance characteristics of the pump, as shown in Figure 6. The pressure head is described as;

$$\text{Pressure head, } H = \frac{\Delta P}{\rho . g}$$

Where ρ= Fluid density ΔP= Pressure difference

The H-Q curve depicts in what way the pump responds to pressure changes. This ability of flow changes in regards to pressure variations is called pressure sensitivity. A flat H-Q curve The high-pressure sensitive pump and low-pressure sensitivity pump attribute to a flat H-Q curve and a steep H-Q curve, respectively. The pressure sensitivity may influence the outcomes in LVAD application. However, this aspect is not completely known (179). The numerical results are compared with the experimentally computed H-Q curve for the data confirmation and vice versa. The variation error values are commonly found to be within 5% when operating at nominal operating conditions, and for other operating conditions, it has been found to be around 10% (68,65,69,71).

"[Insert Figure 6]"

A plastic prototype of axial flow LVAD with a fully magnetic levitated impeller was manufactured based on certain design criteria. The experimental results of this pump were calculated, and a comparative study was done in reference to those of CFD results. The pressure versus flow rate curve for the different rotational speeds of the impeller is shown in Figure 6 (71). Most of the CFD studies and hydraulic performance outputs have indicated that the axial blood pump is generally designed for obtaining an outflow of 5-6 L/min at a pressure head of 100 mmHg (156,157,161,165). Generally, the H-Q curve for axial pumps follows the trend, as shown in Figure 6. The above indicates that a well-validated CFD can be used with confidence to predict flow characteristics in a VAD for design and operation purposes.

5.2 Hydraulic power

The amount of fluid delivered by a pump in a cycle is quantified by Hydraulic power. Hydraulic powerP_h is calculated by the following equation:

$$P_h \ = pgQH$$

Where; H= Pressure Head, ρ= Fluid density, ΔP= Pressure difference

Thus the power requirement can be reasonably well predicted using CFD, and
that can be used in developing system design for the VAD.

5.3 Hydraulic efficiency

The hydraulic efficiency measures the percentage of hydraulic power from the mechanical power transmitted by the shaft of the pump. Thus the hydraulic efficiency of a heart pump is the ratio of energy transferred to the flowing blood in the pump (output) to the energy of the rotating shaft (inlet). The graph between hydraulic efficiency and the volumetric flow rate of axial blood pumps with the variation of vane outlet angles is shown in Figure 7. Sang and Zhou (146) have investigated that the guide vane outlet angle of 10º gives optimum hydraulic efficiency and pressure head. The pump efficiency increases with flow rate up to a point where turbulence intensity is dominant, and it decreases after this point even after the further increase in flow rate. The flow rate corresponding to optimal efficiency is considered as best efficiency point" (BEP). The heart pump should be working at this point to obtain optimal performance.

"[Insert Figure 7]"

A heart pump's efficiency varies from 20-30% (15,69,71,74,77,180). Since the pump runs under widely varying situations, it is preferred that the efficiency should have a minimum fluctuation over the VAD's working range (76), and the nominal design point is generally considered close to the flow rate of a healthy heart (69,74).

The design modifications of VADs alter the BEP. Therefore, designers should take utmost care during design modifications of VADs. The use of the CFD technique in VADs evolution from an engineering point of view is reviewed up to this section. The following sections explore the biological modeling, which helps in quantifying the clinical performance of VADs.

6. Hemolysis

The condition when hemoglobin gets released from red blood cells (RBCs) in the plasma is known as hemolysis. The free hemoglobin is toxic for the kidney if the released amount is above a certain critical level. In the worst situation, it can lead to the failure of multiple organs. Hemolysis occurs due to the deformation and fragmentation of RBCs by shear stress. The approximate range for the shear stress lies between 0.1-50 Pa for a normal human vessel (181). Sublethal damage of the RBCs is the main cause of hemolysis in the blood pump, and catastrophic damage can occur when the shear rates exceed by 42,000 s^{-1} (182,183). The range of shear stress for commercial blood pumps varies from 20-100 N/m^2 with 1 second of exposure time per passage, whereas for small blood pumps, the values may be very high and can increase up to 400 N/m^2 with an exposure time of 1 ms (184). The hemolysis quantification was started by Allen in terms of hemolysis index in 1960 for extracorporeal application (185). A traditional hemolysis index was suggested by Cappelletti *et al.* (186) on the basis of free hemoglobin mass present in every 100 ml of blood. However, the Hematocrit, that is, the ratio of RBCs volume to the total blood volume, cannot be ignored while quantifying hemolysis. The improved traditional hemolysis index with hematocrit taken into consideration is given by the following equation.

$$\text{NIH (g/ 100 l)} = \Delta \text{ free Hb} \times V \times \frac{100-Htc}{100} \times \frac{100}{Q \times T} \tag{1}$$

Where, Δfree Hb, free hemoglobin concentration increment in the inspection timespan (g/l);

V, the system volume; Q, the flow rate of the blood (L/min); Htc, hematocrit; T, inspection timespan.

Mueller et al. (187) have given the modified index of hemolysis (MIH) by considering both hematocrit and hemoglobin concentration. Since the plasma hemoglobin concentration also plays a significant role in altering the blood damage rate. The following equation is a modified version of the earlier equation.

$$\text{NIH (g/100 l)} = \Delta\text{freeHb x V x } \frac{100-\text{Htc}}{100} \text{x} \frac{100}{\text{QxTxHb}} \tag{2}$$

Where Hb, the concentration of the total hemoglobin at the initial time (mg/l), and other symbols denote the same meaning as written above.

Hemolysis and platelet activations rely on the function of shear stress level and time of exposure (188). Hemolysis index (HI) is calculated in percentage based on an empirical formula established by

$$HI = C \times \tau^{\alpha} \times t^{\beta} \tag{3}$$

Where t expresses the exposure time (s), and τ (N/m2) denotes the scalar shear stress tensor. C, α, and β are experimental constants. This empirical relation is more widely accepted for hemolysis prediction than discussed methods above for blood handling devices due to its simplicity (189).

The values of these constants have been determined by two different researchers (188,190). The data acquired by several investigators based on the above empirical relation confirms that both models give the same value of hemolysis in blood pumps (191). The above empirical relation is applicable for steady shear stress only. However, the axial blood pumps are three-dimensional and transient in nature. Thus models suggested by researchers were for calculating the value of one-dimensional shear stress, τ for the above relation by adding components of viscous and Reynolds stress tensor (192).

Many researchers have proposed many empirical relations to quantify hemolysis of blood pumps on the basis of shear stress and exposure time by tracing the particle trajectories (193–198). A novel way of calculating Hemolysis is given by Yanjiao *et al.* (199) based on real-time parameters of the axial flow pump. Most of the blood pumps used for mechanical circulatory support have cannulas. The inlet cannulas of the axial blood pump are mostly not affected by high shear rate or by flow separation (98). However, the cannulas flow fields of the axial blood pump are closely related to the thrombus property (125,136).

In light of the current situation, hemolysis models can be broadly classified in two aspects:

- Strain based or Stress based
- Based on either on Eulerian approach or the Langrangian approach

In stress-based models, the hemolysis is considered at a local instantaneous point as a direct function of the local instantaneous shear stress. Such models employ direct empirical relations shown in equation (3), which is generally in the power-law form. Stress-based models are widely used in most of the literature, whereas strain-based models, although in limited use, consider the strain developed by the shear stress accumulation in the RBCs. In the Langrangian approach, hemolysis is predicted along a trajectory of flow, whereas in the Eulerian approach, it is predicted at a fixed point. The power-law equation model is an empirical model which directly predicts hemolysis via shear stress and exposure time. However, researchers observed that this model overestimates the hemolysis in pumps and other blood-handling devices (194,200).This performed well in the case of moderate constant shear stress and exposure time. Furthermore, experimentally it was observed that below a certain value of shear stress (threshold value), no hemolysis occurs (166,201). Thus for hemolysis prediction, the value is considered to be zero till the threshold value and above that power-law equation is applicable. This sudden change in hemolysis value just after the threshold value creates a discontinuity. The discontinuity can be overcome by introducing a ramp function.

Another derivative of the power-law equation model is the Purely Eulerian Power Law model. This model was derived by rewriting and differentiating it with respect to time for quantification of local hemolysis (198,202–204). Moreover, this model is advantageous over the former model as it can be implemented in a CFD solver. The empirical formulation for the blood pump model is analogous to the power-law equation up to some extent structurally. This model includes the ratio of the pump's volume (inner to whole) and pressure head with exposure time and shear stress (205). This model is valid only for a range of shear stress 1 to 100 Pa, for exposure time 0.1 s to infinity, pressure head 50 to 500 mmHg, and volume ratio 0 to 0.01, as it was derived from experimental data on the basis of regression analysis (206). Lagrangian formulations of the power-law model predict hemolysis based on the estimation of accumulated blood damage strategy. In order to determine accumulated blood damage, two different approaches were applied (207). The first approach to estimate blood damage accumulation, integration of infinitesimally small time steps along pathlines, can be summed up to calculate the blood damage by the formulation proposed by the Grigioni et al. (207). This assumption is based on equation (3). An alternative approach was based on the effect of mechanical dose, which is a function of the destruction of cell membrane occurring in each grid cell along the pathline, and accumulated blood damage was calculated (197,208,209). The major challenge of this model is that it does not consider space steps along a pathline equally. This poses a challenge to inducting this model into the CFD solver. Gu and Smith (210) proposed a Langrangian power-law formulation for the closed-loop circulations model, which is a modified version of the equation (3). This model represents the path lines passing the flow field a number of times. This model predicts correlations only meant for average hemolysis. However, it does have serious limitations. The medical devices used should be calibrated with a broad range of speeds and time exposure values. The viscoelastic Langrangian model is analogous to the spring-damper system, which predicts hemolysis by considering the viscoelastic properties of the RBC membrane. This presented model imitates RBC membrane. Though, it does not directly relates the elementary molecular structure of the cells (211). According to strain-tensor based model, the RBCs' deformation and hemolysis occur due to attainment of previous shear stress and instantaneous deformation respectively (200,212). This model considers the RBCs like a droplet of deformable type during the flow. Its morphology and orientation can be modeled by a symmetric, positive-definite second order tensor. It has two major drawbacks, which are, its complexity and being computationally difficult to solve. Strain-scalar based Lagrangian model, predicts the hemolysis based on a ratio of scalar for area strain of cell membrane s to critical strain for membrane rupture sc (known as hemolysis ratio, s/sc). When the ratio s/sc > 1, complete rupture of travelling RBCs will occur and if the ratio s/sc <= 1 , it predicts hemolysis ratio from local membrane extension (213,214). Though many hemolysis models have been suggested by researchers but the accuracy and validity of such models still is a challenging issue while implementing them with CFD simulations (215,216).

Unfortunately, there are no clear indications of their applicability or limitations. Moreover, there is a lack of proper experimental evidence to validate these models in real situations. Overall, no models are accepted for predicting all kinds of induced hemolysis in different blood handling devices. The above-discussed hemolysis models and their comparisons are enlisted in Table 4. The existing models need further development by emphasizing the experimental evidence and well-documentation that can be more accurate, reliable, and reproducible. In addition, the accuracy of these hemolysis models may be improved by considering blood rheology under different situations such as RBC diffusion, the effect of heating on hemolysis, membrane hardening, and membrane deformation under shear stress.

Table 4: The hemolysis models and their advantages and disadvantages (215).

S. No.	Hemolysis models	Type/ Approach	Advantages	Disadvantages
1.	Power law equation (188)	Stress based, Lagrangian and Eulerian both	Simple in application	Very elementary
	Fully Eulerian power law (198,202)	Stress based, Eulerian approach	Completely Eulerian, time-dependent, repeated hemolysis prediction possible	Very elementary
2.	The empirical formulation for blood pumps (205)	Stress based, Lagrangian approach	Simple in application	Empirical relation, valid for limited application
3.	Lagrangian formulation of the power-law (197)	Stress based, Lagrangian and Eulerian both	Predicts history of shear, based on strain up to some extent	Based on stress in reality

4.	The lagrangian power-law formulation for closed-loop circulation (210)	Stress based, Lagrangian approach	Predicts reduction of hemolysis for repeated passes in circulatory systems	Very elementary, Eulerian form is absent
5.	Viscoelastic Lagrangian model (211)	Stress based, Lagrangian approach	Predicts hemolysis on various time scales	Applicable for constant stress only
6.	Strain tensor base Lagrangian model (212)	Strain based, Lagrangian approach	Predicts relaxation and deformation history	Based on tensor and their eigenvalue operations
7.	Strain scalar base Lagrangian model (213,214)	Strain based, Lagrangian approach	Predicts history of deformation	Incompatible with transport equation, ignorance of shear stress <threshold shear stress

7. Thrombosis

Thrombosis is characterized by the formation of a clot or a thrombus, which is one type of protection mechanism of the body to avoid blood loss. Normally coagulation increases the blood clot to rectify the damaged vessels of a normal human (10). Thrombosis is related to three vital aspects: surface in contact, state of blood cells, and the flow scenario, as shown in Figure 8 (Virchow's Triad) (15,217). Notably, discontinuous patterns of the flow and inflammatory conditions inside the vessels due to hemodynamic, genetic, and dietary parameters could yield unwanted thrombus formation. Hence, inside VADs, areas of high shear stress, recirculation, or stagnation, or blood connecting surfaces having low hemocompatibility must be avoided. Therefore, for the fabrication of the implant, the surface that comes in contact with blood must have high biocompatibility (9,218).

"Insert Supplementary figure here [Figure 8]"

The thrombus development process is initiated due to non-physiological reasons, resulting in pathologically high clotting. If a thrombus develops on the surfaces of the components of the heart pump, it may affect its function, and if the clot flows as an independent thrombus with the bloodstream, then it may block the passage of the various organ. These thromboembolic incidents are the main cause of mortality in heart patients living with implanted VAD. Cavitations, high shear stress, very low shear stress, sharp edges present on the surfaces, high roughness on the surfaces, narrow passage inside the pump, flow stagnation, recirculation, and flow separation region are the main causes of hemolysis, thrombosis, and platelet activation so these parameters should be considered for VAD design (10). Another way to reduce the chances of thrombus formation is by selecting appropriate device material, as some VADs are manufactured or coated with a biocompatible material, such as INCOR LVAD and MicroMed DeBakey VAD (219). The 3rd generation VADs have minimal risk of thrombosis as it is incorporated with magnetic levitation, but in general, this issue remains a major concern of VAD implantation (220). This problem of thrombosis in the blood pumps is a major concern and has been an active area of research for researchers and medical professionals (220,221). The blood pumps basically encounter two types of thrombosis problems, white thrombus, and red thrombus. The white thrombus formation starts due to the shear activation of platelets, whereas red thrombus formation originates due to stagnant blood (220). The mechanically induced shear stress operating on platelets for a longer duration triggers the formation of thrombus, aggregation, and deposition (222).

The CFD technique has been extensively used to model, examine and control the possibilities of occurrence of thrombus formation for the purpose of understanding its process and assessing the hazards in various types of blood pumps. The red thrombus risks can be predicted by qualitative analysis of stagnation points, flow patterns, and non-uniform and recirculation of flow patterns. The frequently used axial pump, HeartMate II, was simulated for the prediction of thrombus formation. These simulation results predicted critical regions of thrombus formation on the basis of recirculation and flow separation zone. The housing upstream of the impeller and diffuser zone was shown to have a high risk of thrombus formation on the basis of results (38). The white thrombus formation risk can be predicted quantitatively on the basis of shear stress acting on the platelets (222). The improved CFD simulations of HearMate II were carried out to predict platelet deposition and thrombosis formation, and the results have shown good agreement with the actual investigations. This study also evaluated that the lower flow rates increase thrombus accumulation (223). The flow pattern within blood pumps may lead to the platelet activation mechanically, which may start thrombus formation. Many researchers suggested various types of models for platelet activation (100,207,224,225). Moreover, various models for platelet aggregation that also incorporate chemical activation were formulated by researchers (226–230).

Blyth and Pozrikidis (231) used the boundary element method in conjunction with adhesion bond dynamics to model a single platelet sticking to a surface. Xu et al. (232) proposed a model to study thrombus formation. The model used Navier-Stokes equations for convection and diffusion of blood flow, while a discrete cellular Potts model was used in conjunction for solving platelets and RBCs-related equations. Wu *et al.* (233) gave the recent thrombosis model, and this high-fidelity model was used for the continuous blood pump, HeartMate II, for thrombus prediction (234). Nevertheless, the flow field in the cannulas is deeply associated with thrombosis characteristic of the axial blood pump (98,125). The reports show that thrombus may block the cannula and passes of the blood pump (235,236). The exact mechanism of thrombosis initialization and its progression are still unknown.

Significant attempts have been made for thrombosis modeling, but the existing models are limited to idealized geometries. The development of an accurate and realistic thrombosis model is very cumbersome and computationally intense. The CFD-based optimization may be used to improve the existing thrombosis models by bridging the gap between experimentally obtained data and computational simulation. Further, the sensor and actuator-based indicator may be created for evaluating the thrombus initiation regions for taking corrective measures as per need. However, this may increase the complexities of pump design. Therefore, thrombosis, hemolysis, and platelet activation should be minimized while designing the blood pump to improve the patient's life quality.

8. Optimization

CFD tools are often used to investigate flow characteristics in blood pumps; however, the best use of CFD for VADs is quantifying the effects of shear stresses on the impeller blades and/or shroud of the pump and enumerating clinical parameters such as hemolysis and thrombosis in the pumps. The clinical aspect of the pumps includes predicting hemolysis and thrombosis risks associated with a design to ensure the success of individual pumps. Typically, Designers of pumps focus on how to use CFD results to improve design and performance. This is usually done using a complicated trial and error method. The CFD results and their uses are used by the designer to improve pump design based on their judgment and experience. The designers develop different designs before choosing which one to carry on with or modify again to obtain targeted results, as shown in Figure 4. The pump components may be designed on the basis of requirements such as expected velocity triangle, specific speed, and other relevant parameters, as mentioned in the study (53). However, the exact performance of a pump or its components may not be predicted with conventional 1-dimensional modeling with high accuracy. The CFD technique may be used for these tests in an efficient manner by using an advanced trial and error method that will help to adapt the best design parameters corresponding to its overall performance. Many research papers are available on design improvements of axial blood pumps, and some are explored here. Zhang et al. (75) eliminated the flow recirculation present in a novel design of axial blood pumps by incrementing the blade expansion angle of the rotor from 0° to 20°. Qi et al. (165) analyzed impeller models with different heights of the blade, such as 1.5 mm, 2 mm, and 2.5 mm, using the CFD technique. The results have shown that the impeller model with a blade height of 2 mm was accepted in terms of clinical requirements. Triep et al. (159) evaluated as well as tested three different impeller geometries for performance enhancement of a micro axial blood pump used for temporary support by using the CFD technique and digital particle image velocimetry. The improved results were obtained in respect of hydraulic efficiency and hemolysis by the impeller that was shown to have minimum drag. Yu and Zhang (237) used CFD modeling to evaluate the hydraulic and hemolysis performance of an axial blood pump which had similar design features as HeartMate II. The distance and clearance between impeller and diffuser were varied from 1 mm to 7 mm and 0 to 0.3 mm, respectively, for better hydraulic performance and low hemolysis. The CFD results predicted that a 2 mm distance and 0.2 mm clearance gave the best results. These are only the cases of performance improvement using the CFD technique. Selishchev and Telyshev (167) optimized the design of sputnik-VAD by reducing the bearing mountings of the diffuser and straightener; hence it reduced size, weight as well as energy utilization. However, in this study, hemolysis and thrombosis were not taken into account.

An optimization technique requires CFD simulations to be integrated with an optimization algorithm that relies on an automated iterative process for design improvement, as shown in Figure 9. This process also needs expertise in a well-adapted iteration loop rather than a traditional design approach (238). The CFD-based optimization faces a major challenge in developing fully automatic and powerful looping (130), and considerable developmental opportunities exist.

"Insert Supplementary figure here [Figure 9]"

Most of the literature available shows only regional optimization without considering the whole parameters of the pump and involves the manual procedure for optimization (145). For example, Zhu et al. (129) carried out optimization of the only diffuser for design improvement of an axial blood pump without considering straightener and impeller parts. Gouskov et al. (239) optimized an axial blood pump considering four geometry parameters such as impeller length, blade angle at the outlet of the impeller, blade angle at the inlet of the diffuser, and bushing diameters using an LP-Tau method as a global optimization algorithm. However, a large part of the process of optimization remained manual. Hai et al. (130) optimized Archimedes screw blood pumps based on an evolutionary algorithm coupled with the CFD technique. However, various assumptions, namely constant shaft diameter, without consideration head and tail of the shaft were introduced. In this study, only seven design variables were considered to optimize the design. In recent literature, Tesch et al. (240) used a fully automated iterative enhancement method for the global optimization of an axial blood pump. This optimization process was based on a modified evolutionary algorithm coupled with the CFD technique. There were fourteen input variables chosen during the optimization process. The optimization results minimized the wall shear stress and hence reduced the hemolysis. However, the straightener was not considered in this optimized axial blood pump.

Generally, gradient-based or derivative-free methods are chosen for the optimization as per the requirement and suitability of the problem. However, the higher number of parameters for the selected design may increase the implementation complexities and computational cost. In optimization of the axial blood pump, each objective function is evaluated by the parameters obtained from CFD analysis/experiments/analytical tools. This task is expensive and requires post-processing steps. The general optimization may be given with the related constraints,

$$\text{Minimize} \qquad J(x)$$

$$\text{Subjected to} \qquad x \in \Omega,$$

Where J: Rn→R refers to a cost function, and x is the vector of design variables. The design variables may be defined by $\Omega=\{x \in Rn\ I \leq x \geq u\}$, where $I \in Rn$ and $u \in Rn$ are vectors of lower and upper bounds on x consequently. The function J(x) is a cost function with subjected constraints, and it depends on the solution of the Navier-Stokes equation and post-processing of simulations that gives the values of the cost function.

The gradient-based optimization techniques have made significant progress in the biological field even after the current challenges of implementation. The optimization of the axial blood pump and its components based on gradient methods has been implemented by various researchers (241–243). However, the complexities of problems often limit the implementation of gradient-based optimization methods (129,244).

The derivative-free optimization methods are the alternative technique that may play a significant role in blood pump design. A popular derivative-free method is named surrogate management framework (SMF), which is based on search optimization technique and depends on surrogate function for efficiency enhancement (245,246). This is the only method based on derivative-free optimization, which has convergence theory (247,248). Generally, the SMF algorithm involves a two-step SEARCH step and POLL step. The SEARCH step uses a Kriging surrogate function for enhanced efficiency, and the POLL step assures the convergence for local minima. The SEARCH step applies the surrogate for choosing the point which enhances the cost function appropriately; however, it doesn't require convergence essentially. The POLL step assures the convergence by selecting trial points close to the current best point on the mesh that is investigated to ensure being the current local optimizer in the mesh (249,250).The implementation of the SMF method is easy and has various advantages, such as overcoming uncertainties in model parameters and design variables in comparison to other methods (16,251). The SMF methods have been used in various application fields such as aerospace, transportation, management, and clinical fields. Moreover, it has also been used for complex engineering designs and problems (252–254).

The SMF method was implemented for the optimization of three representative cardiovascular geometries that are stenosis, vessel bifurcation, and end-to-side anastomosis (250). A fully automated CFD-based surrogate optimization method was used to optimize coronary stent hemodynamically by considering the number of stent cells and optimal strut angle value (255). Yang et al. used the SMF method for design optimization of Y-graft for fontal surgery by considering six design parameters in the optimization algorithm (256). Long et al. used an SMF-based method for design optimization of pediatric pulsatile ventricular assist devices using fluid-structure interaction modeling in the parallel computing environment. This optimized design significantly improved the efficiency compared to the previous one (248). The SMF method has been developed and analyzed for cardiovascular shape optimization with a concurrent evaluation strategy (257). The selection of the most suitable surrogate model is very crucial during optimization, which depends on the complexities of the engineering application as well as accuracy, computational time, and problem size (254). Thus SMF method may play significant role in terms of design optimization of axial blood pumps by considering design parameters with hemolysis and thrombosis taken into consideration in optimization steps. The CFD simulation-based performance optimization of axial blood pumps by taking hemolysis and thrombosis into consideration may reduce the adverse hemodynamic conditions and thrombosis problems during the pump operation.

9. Conclusion and future directions

Some notable conclusions and future directions can be drawn from this rigorous literature review in the area of modeling and simulation of axial blood pumps using CFD as a tool. By incorporating the splitter blade, the current design could be improved. Furthermore, the effect on hubless design has not been thoroughly investigated, which could be studied for further improvement in hydraulic and clinical performance rather than conventional design. Nonetheless, the combination of both aspects has not been studied, which could improve the pump's overall performance. Furthermore, the effect of blade profiles on the pump's hematological performance is unknown. As a result, these aspects can be explored for performance enhancement. Furthermore, the simulation results can be improved by taking into account the rheology of blood-pump interactions. The most recent

axial blood pumps have made significant advancements over their predecessors. The size of the axial blood pump can be reduced further in the future, which will be very beneficial for both implantation and operation. The smaller size of the axial blood pump implant will consume lesser power than older models. Furthermore, the impeller's constant rotational velocity may be replaced by speed modulation with an appropriate profile to mimic pulsatility similar to that of a normal heart. This can be accomplished by incorporating a suitable speed modulation algorithm into the pump design. For optimum performance, the effect of different speed modulation profiles should be computationally studied before applying it to the algorithm. Furthermore, the airfoil shape of the impeller and diffuser may be optimized to improve hydraulic performance while also reducing the overall size of the blood pump. Because of the computational cost involved, mathematical and turbulence modeling of actual blood flow in veins and the heart remains elusive. As a result, the risk of thrombosis and hemolysis must be assessed using well-validated and verified models. A better correlation between simulation and optimization techniques is required to improve the design aspects of blood pumps while taking errors, uncertainties, and constraints into account. With the introduction of computationally simulated new axial blood pump designs, the need for axial blood pump prototypes will emerge for practical purposes. After the simulation process is completed, validation and comparison of thrombosis and hemolysis data with simulation results and actual calculated data on patients and animals will be required. Finally, while developing the model, the physiological requirements of the patients should be kept in mind. The advanced simulation CFD tool would be able to precisely predict the hemolysis. The innovative design of the axial blood pumps through blade profile selection, design modifications, impeller rotational profile selection, and size miniaturization, combined with CFD-based optimization, will result in reduced thrombosis and, as a result,

increased blood pump life. Furthermore, by incorporating lumped parameter models, there is a greater need to develop a strategy for automatically regulating the impeller rotational speed based on the patient's physiological requirements. Additionally, researchers developed suitable levitation stability of passive magnetic bearings for nutating disc pumps and examined their viability for VAD application using mathematical models, experimental testing, and CFD simulation. More experimental and computational studies are needed to gain a precise understanding of performance parameters such as efficiency, durability, motor control, and other design aspects that may aid in creating a better version of the pump.

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