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Axial and centrifugal continuous-flow rotary pumps: A translation from pump mechanics to clinical practice

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The recent success of continuous-flow circulatory support devices has led to the growing acceptance of these devices as a viable therapeutic option for end-stage heart failure patients who are not responsive to current pharmacologic and electrophysiologic therapies. This article defines and clarifies the major classification of these pumps as axial or centrifugal continuous-flow devices by discussing the difference in their inherent mechanics and describing how these features translate clinically to pump selection and patient management issues. Axial vs centrifugal pump and bearing design, theory of operation, hydrodynamic performance, and current vs flow relationships are discussed. A review of axial vs centrifugal physiology, pre-load and after-load sensitivity, flow pulsatility, and issues related to automatic physiologic control and suction prevention algorithms is offered. Reliability and biocompatibility of the two types of pumps are reviewed from the perspectives of mechanical wear, implant life, hemolysis, and pump deposition. Finally, a glimpse into the future of continuous-flow technologies is presented.

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The evolution of mechanical circulatory support from volume-displacement pulsatile pumps to continuous-flow (CF) rotary pumps has ushered in a new era for treatment of end-stage heart failure. Increasing use of these pumps in the clinical arena is related to multiple positive attributes of this class of pumps, including their smaller size, improved durability, and enhanced survival with less morbidity.1,2 As acceptance of these classes of pumps continues to grow and newer-generation pumps are developed, a complete understanding of their mechanics and the inherent differences between centrifugal and axial-flow pumps becomes imperative for patient care and decisions regarding pump selection. In this review, we describe the basic engineering differences between the 2 types of CF pumps and provide a description of how these features translate clinically to patient management.

Axial vs centrifugal CF pump design and theory of operation

CF rotary pumps generally consist of blood inlet and outlet ports and a single rotating element that imparts energy to the blood to increase arterial blood flow and pressure. The rigid stationary housing(s) that surrounds and/or lies in the center of the rotating element incorporates some combination of motor windings, permanent magnets, electromagnets, or
mechanical bearing surfaces that act to drive and support the rotating element.

The primary difference between centrifugal-flow and axial-flow pumps lies in the design of their rotating elements (Table 1). When one considers the theory of operation of a centrifugal CF pump, its rotating element acts as a spinning disk with blades that can be viewed as a “thrower,” meaning that the fluid is captured and thrown tangentially out off the blade tips. In contrast, axial CF pump rotating elements operate like a propeller in a pipe and can be viewed as a “pusher.” This mechanism can also be viewed as an “auger” trying to screw itself into the inlet fluid, against the “resistance force” at the outlet, to overcome the difference between pre-load and after-load.

### Bearing design

A further distinction made between CF pumps is the method used to support the rotating element (rotor). Listed below and illustrated in Figure 1 are several different kinds of bearing elements that are used to accomplish this:

- **Mechanical/pivot:** Rotor is suspended with mechanical bearings on spherical surfaces rotating in sockets (eg, HeartMate II, Thoratec Corporation, Pleasanton, CA).
- **Hydrodynamic:** Rotor derives lifting from fluid forces in thin, fluid blood films separating rotor and pump housing based on the relative motion of surfaces (eg, HeartWare HVAD, HeartWare International Inc, Framingham, MA).
- **Electromagnet/position sensor:** Suspend rotor using electronic position control and electromagnets (eg, DuraHeart, Terumo Heart Inc, Ann Arbor, MI; and HeartMate III, Thoratec Corp).
- **Permanent magnet:** Repelling magnets in rotor and pump housing suspend the rotor. Permanent magnets alone cannot suspend a rotating element because magnetic forces change continuously with the position.
Figure 1  Rotary pump bearing types. (A) Mechanical pivot bearing: a drawing of a generic axial-flow pump illustrates how the rotating element (rotor) is suspended by pivot bearings on either end as it spins. These bearings, which are usually made from very hard jewel or ceramic materials, are in mechanical contact while the rotor is spinning. (B) Electromagnetic bearing: Illustration shows electromagnetic levitation of the INCOR (Berlin Heart GmbH) axial continuous-flow pump rotor. Levitating electromagnets act on opposing permanent magnets (black) on both ends of the rotating element to suspend the rotor axially, while the permanent follower magnets in the rotating element and electromagnetic forces induced in the motor stator suspend the rotating element radially. The arrows show the directions of magnetic forces. (C) Hydrodynamic radial bearing: Diagram shows a cross-section through a generic rotating cylinder (white disc in cross-section) within a stationary cylinder that is suspended by thin fluid film pressures (fluid is grey in the diagram). This bearing has no mechanical contact because the hydrodynamic forces keeping the rotating element away from the stationary element increase as the distance between them decreases. The magnitude and direction of these fluid forces is represented by the height and direction of the arrows. (D) Hydrodynamic thrust bearing: Diagram shows the cross-section of a generic thrust bearing, consisting of 2 plates, where thin fluid film pressure acts to separate a rotating surface (2) moving to the right in this figure relative to a stationary surface (3). These forces are created by the fluid wedge created between geometry of the moving plate relative to the stationary plate and again are represented by the direction and height of the arrow in the diagram. An example of this bearing type is shown in Panel E. (E) Hydrodynamic thrust and permanent magnet bearing combination: Assembled component view of the HeartWare HVAD centrifugal continuous flow LVAD on the right and cross-section on the left. Permanent magnets (1-1) align and separate the rotating disc impeller (2) from the conical stationary center shaft of the pump housing to form a permanent magnet bearing. This bearing is combined with thin film fluid pressures generated at the tapered hydrodynamic thrust bearing surface of the rotating disc impeller (2) that creates the fluid wedge needed to create sufficient hydrodynamic forces to lift the impeller off the stationary housing bearing surface (3). The red line between the rotor (2) and housing (3) is the blood fluid film.
of the rotor. Therefore, permanent magnets are generally used in combination with one of the other bearing types. In a magnetically suspended pump, stabilization in one or more directions of motion is achieved by using hydrodynamic or electromagnetic bearing elements.

As a practical matter, the primary difference in blood pump bearings lies in their complexity and reliability. Mechanical bearings have tiny and precise ceramic components whose advantage is that they position the rotating assembly in all 6 directions of motion and are stable at all speeds and operating conditions. Although their small size limits the surface area exposed to mechanical contact and friction, these surfaces are potential sites for thrombus/fibrin deposition, which is not present in other bearing designs that completely suspend the rotor. The concentration of hydraulic loads on the rotor at these small pivot bearings also makes them theoretically life-limiting for wear and fragile with respect to impact.

Electromagnetic rotor positioning generates real-time electromagnetic forces on the rotor to actively counteract the surrounding fluid and magnetic forces that could cause unstable rotor operation. Among the advantages of this approach are that there is no contact between the bearing surfaces, so there is no life-limiting wear, and that it provides the lowest shear stress to the blood because of relatively large clearances between the rotating element and the housing. However, these systems are complex, requiring position sensors, electromagnets and extra conductors, connector pins, and electronics to execute the dedicated position control algorithm. If, for example, there is an electrical contact failure in a connector pin, or momentary instability encountered in the control algorithm, pump failure can occur. To make the system fail-safe, electromagnetic bearing elements are frequently backed up with hydrodynamic bearings, which require no control.

Hydrodynamic bearings similarly produce no life-limiting contact between the bearing surfaces, but in contrast, are simple and reliable. Once the rotor starts spinning, it essentially “water skis” on a film of blood, deriving lift and separation from its own motion. However, the load-bearing blood fluid film is prone to higher shear stress, and theoretically, more hemolysis can occur. Another design issue is that the rotor and housing surfaces are in contact at startup and shutdown, when there is no relative motion. The bearing surface material properties need to accommodate this friction to avoid damage at these 2 times. Surface coatings, if used, require extended endurance testing to verify multiyear reliability.

### Hydrodynamic performance

The pressure across the inlet and outlet of any hydrodynamic pump is termed by engineers as the “pump delta P” or “head pressure.” Figure 2A shows the typical pressure difference between the left ventricle (LV; pump inlet) and aorta (pump outlet) of a failing heart supported by a rotary pump. Figure 2B and C compare the typical hydrodynamic performance curves (pump head curves) for centrifugal vs axial CF pumps, respectively. A pump head curve compares the relationship between pump flow and pressure difference across the pump ports (delta P) at one operating pump speed. Figure 3 shows a full set of pump head curves over the full range of operating pump speeds for the axial HeartMate II and the centrifugal HeartWare HVAD and Terumo DuraHeart left ventricular assist device (LVAD).
Most important to note is that most centrifugal pumps have what is called a flat head curve, where they operate over a very wide range of flows for a very small change in delta P across the pump. The example in Figure 3 shows that for 1 cardiac cycle in which pump delta P swings from 40 to 80 mm Hg, centrifugal pumps have a very large swing in flow (0 to 10 liters/min), acting almost like a pulsatile pump with high peak systolic flows and low, even negative, diastolic flows. One can think of a centrifugal pump with a flat head curve as an on/off CF pump cycling between high-flow and low-flow as its output surges with the beating of the ventricle. This creates inherent high pump flow pulsatility in response to changing LV pressures (Figure 2D). In contrast, a typical axial-flow pump has a steep head curve where there is a linearly related increase and decrease in flow with decreasing and increasing pump delta P. In this example, the 40- to 80-mm Hg swing across the pump conduits produces less flow pulsatility, ranging from 3 to 7 liters/min during a cardiac cycle. As the text that follows explains, this difference in pump-flow pulsatility affects the diagnostic and control feedback available to the respective pump controllers. Figure 3B and C demonstrate that the actual degree of flatness of centrifugal pump head curves can vary with design, with the HeartWare HVAD being less flat than the Terumo DuraHeart design.

**Inlet cannula suction and control of pump operating speed**

The likelihood of high inlet cannula suction in axial vs centrifugal CF pumps is related to the flow vs delta P relationship described above. A CF pump is blind to the absolute value of the pressures at the inlet and outlet ports and responds only to the total differential pressure across the pump. At any given pump speed, a CF pump would have the same flow at inlet and outlet pressure values of 0/100, 100/200, and –50/50 mm Hg, because it sees only the delta P of 100 mm Hg. This has clinical significance during a low LV volume state in which pump flow decreases, such as one might expect during conditions related to hypovolemia associated with right ventricular (RV) failure or bleeding. As can be seen from the head curves in Figure 2B, a
centrifugal LVAD with a flat head curve has a fixed differential pressure as flows decrease from 5 to 0 liters/min, so no significant increase in inlet suction occurs at low flows or even total occlusion. However, as a result of the steep head curve characteristic of axial pumps, the increasing pressure differential generated across them as flow decreases translates into much higher suction developed at the inlet during conditions of low LV volume or inlet cannula obstruction. This means axial pumps pull the hardest at the lowest flow, creating the potential for a self-latching condition wherein the pump continues to increase suction as flows fall, resulting in the ventricular wall being sucked in around the inlet. Therefore, centrifugal pumps have an advantage over axial pumps in the avoidance of ventricular suck down at low flow.

A complication of intermittent LV suck down is ventricular arrhythmias. Morshuis et al reported a 65% lower prevalence of ventricular arrhythmia than reported for axial flow LVADs in their clinical experience with the DuraHeart centrifugal pump. Another pathologic condition that higher LV inlet suction can create is leftward shift of the interventricular septum. This not only negatively alters the function of the septal leaflet of this valve, also increase tricuspid valve regurgitation secondary to the mechanics of the septal contribution to RV output but can also increase tricuspid valve regurgitation secondary to the anatomic connection of the septal leaflet of this valve, causing it to be “pulled” away.

Recent trends in patient management have been to operate CF LVADs at lower speeds and at less than the maximum level of cardiac support to ensure continuous or intermittent opening of the aortic valve during operation. This is to prevent reported complications of aortic valve fusion and reduce the potential for aortic regurgitation, but it also reduces the likelihood of suction events.

Flow estimation accuracy based on flow vs power relationships

Centrifugal pumps have a linear current-to-flow relationship across the full range of operating pump flows. In addition, the characteristic current and flow pulsatility in response to the pressures generated across the pump makes the centrifugal pump motor current an accurate sensorless index of pump flow and a sensitive virtual index of the LV pressures during the cardiac cycle. This information allows centrifugal pump controllers to accurately monitor pump flow and the degree of LV unloading by simply monitoring the motor current or power.

In contrast, the correlation between flow and current in axial pumps is not nearly linear over the full operating flow range, is not as well defined as in centrifugal pumps, and offers less accuracy for flow estimation. For example, the HeartMate II axial flow pump does not display flows below 3.0 liters/min. In an intraoperative study of estimated flow accuracy in 20 HeartMate II patients, Slaughter et al concluded that the device’s estimated flow values can be used only “to provide directional information for trend purposes rather than absolute values of pump flow.” The MicroMed DeBakey (MicroMed, Houston, TX) axial pump relies on an ultrasonic flow probe incorporated into its outlet conduit for flow determination. The centrifugal HeartWare HVAD and Terumo DuraHeart LVAD flow estimations derived from motor power and speed inputs have been generally considered accurate and reliable enough to use in clinical assessment.

Accurate flow estimation for axial and centrifugal rotary pumps also depends on the viscosity of blood, which can be estimated from the patient’s hematocrit. Currently, the HeartWare HVAD has a programmable hematocrit setting, whereas the HeartMate II does not. Implementation of automatic control algorithms is expected to be more difficult in axial pumps without direct flow measurements than in centrifugal pumps.

Axial vs centrifugal physiology and control

Pre-load Sensitivity

Cardiac output in the natural heart is determined by the interaction of after-load, myocardial contractility, heart rate, pre-load sensitivity, and LV compliance. Pre-load sensitivity, as it relates to mechanical circulatory assist devices, mimics the relationship between LV filling pressures and ventricular stroke volume defined by the Frank-Starling curves. It is calculated from the ratio of pump output to pump filling pressures at the pump inlet in liters/min/mm Hg. Salamonsen et al reported an average pre-load sensitivity for the DuraHeart, HeartWare HVAD, HeartMate II, and INCOR (Berlin Heart AG, Berlin Germany) CF rotary pumps of 0.105 ± 0.096 liters/min/mm Hg, which is almost 3 times lower than the 0.275 liters/min/mm Hg reported for the human heart. The limited pre-load sensitivity of CF devices theoretically limits their rate of increased output in response to increasing LV venous return. There is no evidence in the literature to suggest that axial or centrifugal pumps would have any significant clinical difference in pre-load sensitivity.

Reports of greater LV volume unloading by pulsatile vs CF LVAD pumps may be a consequence of the low pre-load sensitivity of the CF LVADs. Interestingly, a review of all bridge-to-recovery cases at the German Heart Institute from 1992 to 2009 showed that patients with a pulsatile-flow LVAD had an almost 3-fold chance for myocardial recovery compared with those who received CF devices. This may be due to greater volume unloading of the LV by pulsatile pumps. To date, no published reports have compared the exercise tolerance of patients with axial vs centrifugal CF LVADs.

After-load sensitivity

Data from Salamonsen et al on after-load sensitivity of CF rotary pumps support the general understanding that these pumps have higher after-load sensitivity (0.09 ± 0.034 liters/min/mm Hg) than the human heart (0.03 ± 0.01 liters/min/mm Hg). This high after-load sensitivity creates the need to control systemic vascular resistance (SVR) in these patients to guarantee sustained outputs.
Typically, the targeted mean systemic arterial pressures for CF pump patients is 70 to 90 mm Hg, with pressures exceeding 90 mm Hg to be avoided.18 Although axial and centrifugal CF pumps share this generalized categorization, analysis of their head curves suggests some relative differences that can have significant clinical implications. Because centrifugal pumps operate on a flatter head curve than axial pumps, they demonstrate larger changes in flow for any given change in pressure across the pump. If SVR were to increase (increase in delta P), the pump outlet pressure increase would produce an instantaneous drop in pump flow to maintain a constant pump outlet pressure, which means that centrifugal pumps cannot pump against a high blood pressure, thereby causing a lower flow condition.

In contrast, the steeper head curve of axial pumps responds to an increase in SVR by increasing the pressure generated across the pump ports, limiting the decrease in flow by increasing outlet pressure. At low-flow conditions, this retains the capability to enforce high blood pressures with adequate LV volume. However, it also results in high inlet suction that in a low LV volume state can potentiate arrhythmias, create suction events, or lead to hemolysis.

Pump flow pulsatility and automatic physiologic pump control

The degree of pulsatility in the pump flow waveform during a cardiac cycle for CF rotary pumps is termed the flow pulsatility and is quantified in various ways for different pump systems as a pulsatility index (PI). It is typically displayed as some form of the ratio of the peak flow during systole (flow_{max}) or the total swing in flow during a cardiac cycle (flow_{max} – flow_{min}) over the average pump flow (flow_{avg}).3,18,19 This is normally derived by averaging the recorded instantaneous peak flows or flow swings over each cardiac cycle for approximately 10 to 15 seconds and then calculating the average total pump flow over that same period. The formula for calculating a PI for the HeartMate II (PI_{HM II})20 is \( \text{PI}_{HM \ II} = \left( \frac{\text{flow}_{max} - \text{flow}_{min}}{\text{flow}_{avg}} \right) \times 10. \)

For example, if rotary pump flow values for flow_{max}, flow_{min}, and flow_{avg} were 8, 4, and 6 liters/min, respectively, the PI_{HM II} would be 6.7. Multiply the PI_{HM II} by 10, and it can be interpreted as the percentage of the average flow that the pump flow swings through during a cardiac cycle, which for this example would be a 4 liters/min swing or 67% of the average flow of 6 liters/min. The average PI_{HM II} for the axial HeartMate II bridge-to-transplantation (BTT) trial was 5.0 ± 0.9, and the average PI calculated as PI_{HM II} for the centrifugal CorAide European clinical trial was 12.1 ± 2.6, demonstrating more than twice the flow pulsatility.3,18,19 Although flow pulsatility data for the HeartWare HVAD clinical trial are not available, it is recommended to maintain a minimum of 2 to 4 liters/min for pump flow operating ranges of 3.5 to 7.0 liters/min, which equates to a PI_{HM II} value of 5.7.

The significance of flow pulsatility for rotary blood pumps lies in the fact that the degree of flow pulsatility is inversely related to the degree of LV unloading by the pump and directly proportional to the strength of LV contraction and as such can be used as a measure of the LV function under VAD support. Any significant decrease in pump flow pulsatility without a change in pump speed should be investigated clinically for causes of decreasing LV pressures during LVAD support. This typically includes decreasing LV contractility or low LV volume states caused by right heart failure or dehydration. Reasons for the flow PI to increase, without a change in pump speed, include an increase in LV contractility via inotropes, myocardial recovery, and exercise or increased pre-load Starling effects.

It is not surprising that automatic physiologic pump control algorithms have not been used to any significant extent in the clinical axial-flow pumps given their previously stated poor flow estimation accuracy, lower flow pulsatility, and the increased danger of LV suction events. They are typically operated only in fixed-speed mode with a suction-detection control algorithm. The CorAide centrifugal LVAD used sensorless flow estimates, high pump flow pulsatility response to changes in the status of the LV, and much lower LV suction levels during transient periods of low LV volume to successfully implement a physiologic automatic pump control algorithm in its European clinical trial.19,21 In this limited trial of 21 patients, 72% of the total trial implant duration (16.6 patient-years) was run in automatic mode (personal unpublished data).

Although the other centrifugal CF pump systems can theoretically implement similar physiologic automatic control schemes, there have been no reports to our knowledge of any other axial or centrifugal CF pump system that routinely uses a physiologic control algorithm. For the centrifugal CF pump systems, this is most likely attributable to the excellent results to date using fixed-speed mode of operation. Any potential advantages to physiologic automatic control algorithms in CF LVADs will have to wait for future control system developments in the newer CF pump systems.

LV suction-detection algorithms

Centrifugal and axial CF pump systems use some mechanism to detect a suction event; however, the reliability of the centrifugal pump flow calculations at low-flow levels and their higher flow pulsatility levels allow these systems to implement pre-suction avoidance before suction occurs by dropping pump speed when flow PI drops below pre-programmed lower limits. Because rotary pump suction events typically occur during early ventricular diastole, during the lowest LV volumes and pressures, the centrifugal HeartWare HVAD pre-suction detection system looks for a sudden decrease in the pump’s diastolic minimum flow level. The baseline diastolic minimum flow level is continuously averaged and recalculated every 2 seconds, and a pre-suction detection alarm occurs if the baseline flow decreases by more than 40% for 10 seconds.3 Only an alarm is activated, because there is no automatic response to decreased speed under LV pre-suction detection conditions.
Because axial CF pumps typically do not derive accurate pump flow values and do not have the degree of pump flow pulsatility of centrifugal CF pumps, they tend to rely on detection of a suction event after it occurs based on the sensed disturbed flow and motor current feedback when intermittent inlet cannula suction takes place. 22–23 The HeartMate II system does automatically decrease pump speed in response to a suction event. Hertzer et al24 have reported that the INCOR axial flow pump has an automatic anti-suction algorithm based on flow pulsatility falling below a pre-determined minimum level, and recent publications indicate that Berlin Heart Inc, is developing and testing algorithms for automatic pre-load–sensitive control of rotary blood pumps. 25

**Axial vs centrifugal reliability and biocompatibility**

**Mechanical wear and implant life**

With CF LVADs now being used as permanent destination therapy for end-stage heart failure patients, high reliability out to 5 years is generally considered a minimum requirement for permanent use, and a 10-year device is a targeted goal for newer device designs. Most centrifugal CF LVADs are defined as third-generation pumps in which the pump rotating element within the blood flow path is totally suspended during operation by some combination of hydrodynamic and/or magnetic (passive or active) forces, giving these pumps no life-limiting mechanical loads.

Most axial CF pumps use second-generation rotating assemblies that are pivoted on small bearing surfaces in the blood flow path. An exception is the Berlin Heart INCOR axial CF LVAD, which uses magnetic bearings to actively suspend the rotating assembly. Bearing contact points in principle are pre-disposed to mechanical wear and heat generation, limiting the operating life of the pump. Despite the stated theoretic limits to pivoted bearing pumps, the Jarvik 2000 (Jarvik Heart Inc, New York, NY)26 and the HeartMate II axial rotary pumps have both shown excellent reliability, with no reported life-limiting wear of their blood-immersed bearings. 27

However, the German Heart Institute reported in 2010 that of 65 Berlin Heart INCOR and 18 MicroMed DeBakey axial rotary pumps implanted between 1991 and 2009, 3 stopped because of technical failure and 2 Berlin Heart INCOR devices had bearing problems. 28 Interestingly, the HeartMate II axial pivoted-bearing CF pump was designed as a 5-year device, whereas the HeartMate III centrifugal CF pump, now in pre-clinical animal testing, has a stated design goal of a 10-year operating life. 29

Another potential failure mode for the small pivoted bearings in axial CF pumps is susceptibility to shock loads, such as in a car accident, in which high decelerating forces are imposed. These forces are concentrated over very small bearing surface areas and can lead to catastrophic failure. In contrast, most centrifugal pumps have relatively stiff permanent magnet suspension systems to prevent significant axial displacement of their rotors and large hydrodynamic bearing surfaces capable of safely distributing transient touchdown forces due to shock loads. The HeartMate III centrifugal electromagnetic rotor levitation system was designed to withstand accelerations of more than 7 times that of gravity. 29

As is expected in any initial trials of new technology, the later-arriving centrifugal CF rotary pumps identified reliability issues not identified in pre-clinical testing. In the HeartWare HVAD BTT clinical trial, Strueber et al30 reported no device failure; however, 4% (2 of 50) of the devices were electively exchanged because of what was determined to be manufacturing variability of the pumps’ hydrodynamic thrust bearings, resulting in an area of reduced flow that may have been attributable to thrombus formation. After resolving the manufacturing process problem, they reported no further events occurring in more than 450 additional implants. 30

The initial European experience with the DuraHeart centrifugal CF rotary LVAD in 68 patients reported no pump mechanical failures; however, the pumps in 2 patients were replaced electively because of temporary flow interruption identified by the manufacturer caused by a motor manufacturing process that has since been corrected. 31 As experience with longer LVAD support duration accumulates, the answer to the relative reliability of axial vs centrifugal rotary blood pumps will need to play itself out in the clinical arena despite the theoretic advantages offered by third-generation pumps.

**Hemolysis**

Pump hemolysis is directly related not only to pump speed and areas of high shear but also to the residence time of blood in the high fluid shear areas of the pumping elements or bearings. Axial pumps generally have significantly smaller clearances at the impeller blade tips and higher speeds at the outer edge of their blades, creating higher shear levels than for the centrifugal pumps; however, the high shear around the blade edges tends to be relatively short in duration, helping to mitigate this blood damage. Hydrodynamic bearing design is tricky because of the significant residence time in the high shear fluid film that suspends the rotating assembly. However, hydrodynamic bearing hemolysis has been mitigated to low levels by proper bearing design, as demonstrated by the HeartWare HVAD. The current literature does not definitively support any difference in hemolysis rates between the current axial vs centrifugal CF pumps, nor has hemolysis been a significant complication for CF pumps in general; however, the literature indicates that axial pumps may produce mild subclinical hemolysis whereas the newer centrifugal pumps do not.

A 2011 report on implantation of 2 HeartWare HVAD pumps for biventricular support by the German Heart Institute reported no significant increase in free hemoglobin, bilirubin, or lactate dehydrogenase (LDH) blood levels during the first 6 post-operative months. 32 The initial European experience with the DuraHeart centrifugal CF
rotary LVAD showed a decrease in total bilirubin, glutamic-oxaloacetic transaminase, glutamic-pyruvic transaminase, free hemoglobin, and LDH from baseline values for those patients reaching 6 months’ duration, indicating no clinically significant hemolysis.31

The reported incidence of hemolysis for the HeartMate II axial rotary pump is 0.06, 0.02, and 0.13 events per patient-year in its BTT U.S., destination therapy U.S., and European multicenter clinical trials, respectively.33 In a study comparing hemolysis indices in CF LVAD patients supported by the axial CF rotary pump (HeartMate II) and the (VentAssist), centriugal CF rotary pump, patients with the centrifugal LVAD displayed normal haptoglobin levels, low plasma free hemoglobin, and moderately increased LDH. In comparison, the HeartMate II patients showed lower haptoglobin and higher free hemoglobin and LDH values. Hemolysis indices reported by Hetzer et al24 for the initial experience with the Berlin Heart INCOR axial rotary pump showed a statistically significant decrease in LDH to below baseline followed out to 6 months. Plasma free hemoglobin did not increase statistically, but they reported some persistent elevation from baseline in the study. Westaby et al37 documented no significant change in free hemoglobin or LDH for 4 Jarvik 2000 patients, whereas Siegenthaler et al38 reported mild hemolysis not requiring transfusion, stable hemoglobin levels, increased LDH levels, and low haptoglobin levels in 3 Jarvik 2000 patients. Despite these reports of low hemolysis based on pump design, new-onset hemolysis is often the first manifestation of problems in CF LVADs that lead to turbulent flow such as can be seen with a kinked outflow graft or partial pump thrombosis.

Pump deposition

From a perspective of pump design, mechanical wear and heat generation at the bearing contact points in the second-generation axial pumps increase the chances of fibrin deposition and thrombus forming in the blood flow path. All CF pumps are designed around an operating point being a single-flow and pump-speed condition; however, centrifugal pumps can be less sensitive to off design conditions because they do not have stator vanes. Axial pumps have stator (stationary) vanes at the inlet and outlet that guide the flow at the inlet to prevent pre-rotation of the blood as it enters the pump and catch or “scoop up” and straighten the flow at the outlet. The geometry of the vanes is just right for only 1 value of flow/speed (eg, 5 liters/min/10,000 rpm), representing the “design point” for the device. At this point, flow separation from the surfaces (meaning areas of poor wash) is minimized. At conditions of high flow/low speed (low SVR) or low flow/high speed (high SVR), these stator vanes are off design and may accumulate thrombus in the zones of flow separation.

A more aggressive anti-coagulation and anti-platelet therapy was used initially for many of the axial CF rotary pumps (international normalized ratio [INR] 2.5 to 3.5) compared with the HeartMate I pulsatile devices.39 With increased experience in both axial and centrifugal rotary blood pump application, anti-coagulation requirements have decreased for some of these pumps. Analysis of the monthly INR of a 331-patient sub-set of patients successfully discharged in the BTT arm of the U.S. HeartMate II pivotal trial recommends an INR of 1.5 to 2.5 for the HeartMate II to retain a low incidence of thromboembolic events while minimizing hemorrhagic events.40 El-Banayosy et al41 used a more moderate anti-coagulation protocol to manage all rotary blood pumps. They report successfully using a target INR of 2.0 to 2.5 and aspirin at 100 mg daily to anti-coagulate 6 patients with HeartMate II axial, 10 with DuraHeart, and 8 with VentAssist centrifugal rotary pumps, with no device thrombosis, no hemorrhagic or ischemic stroke, and no other thromboembolic complications.31

We must note that the recommendations for maintaining INR in the range of 1.5 to 2.5 to balance the dangers of thromboembolic events with bleeding events have not been confirmed prospectively to date. The lack of mechanical contact, lower bearing shear levels, and use of more biocompatible pump surfaces in the electromagnetically suspended centrifugal pumps may allow operation at even lower anti-coagulation regimens. Use of only anti-platelet agents is being proposed for the HeartMate III centrifugal LVAD now in pre-clinical testing.29

The reported ischemic and hemorrhagic stroke event rates per year for the HeartWare HVAD were 0.11 and 0.05, respectively, in the BTT clinical trial vs 0.09 and 0.05 for the HeartMate II clinical trial and 0.05 and 0.01 in a post-approval study.42 The neurologic complication rate for the DuraHeart centrifugal rotary pump improved after the anticoagulation regimen was decreased from an INR of 2.5–3.5 to 2.0–2.5 in the European clinical experience; however, the incidence of transient ischemic attacks (0.23) was somewhat higher than for the HeartMate II and HVAD pumps (31).

Future directions

The excellent reliability and biocompatibility of the HeartMate II axial-flow LVAD has been widely published. This review, however, suggests that centrifugal rotary CF pumps may have some theoretic and real advantages over axial CF pumps (Table 2). Going forward, the pendulum may swing to this pump design in the continuum of CF rotary pump design and development. It is not a coincidence that the HeartMate III centrifugal CF rotary pump eliminates many of the limitations of axial pump design, including (1) bearing-less low shear magnetic levitation of the pump rotor to increase reliability and durability, (2) an accurate sensor-less flow estimator, (3) the ability to withstand high shock loads, (4) incorporation of a pulse mode to increase the level of pulsatility and even produce physiologic pulse pressures, and (5) the potential for reduction of anti-coagulation requirements to only daily anti-platelet medications. This, combined with the capability to implement physiologic control algorithms based on accurate flow estimates and the inherent advantages of flat centrifugal pump head curves over steep axial pump head curves, allows (1) much lower inlet suction at low flows; (2) lower outlet pressures at high
SVRs, preventing the creation of exceedingly high systemic arterial pressures; (3) higher flow pulsatility that is more sensitive to LV pressure swings, providing both a more sensitive pre-suction detection capability and increased diagnostic feedback as to the functional status of the LV; and (4) safer long-term management of pump speed and pump output optimization. The lower inlet suction at low flows demonstrated by centrifugal rotary blood pumps may be one of the first clinical criteria to be used in selecting centrifugal pumps rather than axial pumps for some patients in diastolic heart failure. VAD candidates presenting with hypertrophic ventricles with small end-diastolic volumes may be less likely to experience arrhythmias due to contact of the inlet cannula with the LV wall when a centrifugal rotary pump is used.

HeartWare International is currently in pre-clinical testing of its next-generation device, the MVAD, which is a wide-blade axial CF pump.43 Whereas the Berlin Heart INCOR rotary LVAD has been designed as a third-generation axial pump using active electromagnetic suspension of the rotor, the MVAD features a magnetic axial rotor and uses a combination of passive permanent magnets and hydrodynamic bearing surfaces to totally suspend the rotor in operation with a displacement volume of only 15 cm³. This is accomplished by a unique rotor with impeller blades that are wider than the blood flow channel between them, affording sufficient surface area for passive hydrodynamic radial rotor suspension. This eliminates the mechanical pivot bearings typical of axial pumps and significantly reduces the pump size by using passive rotor suspension rather than the more complex and larger active electromagnetic bearing designs. Also not typical of axial pump design is an outflow port angled 90° from the pump housing inlet port, similar to centrifugal pump designs. Various versions of the MVAD are being tested within the LV and at the LV apex in animal studies.

Not addressed in this report is the growing interest in miniaturized intraventricular and intravascular CF rotary pumps, which have been and are expected to be primarily axial-flow pumps. Centrifugal pumps are dependent on

Table 2  Summary of Key Physiologic Differences in Axial and Centrifugal Continuous Flow Pumps

<table>
<thead>
<tr>
<th>Pump characteristics</th>
<th>C vs A</th>
<th>Qualitative pump comparisons</th>
</tr>
</thead>
<tbody>
<tr>
<td>Flow pulsatility</td>
<td>C &gt; A</td>
<td>Centrifugal pumps have significantly higher flow pulsatility.</td>
</tr>
<tr>
<td>Estimated flow accuracy</td>
<td>C &gt; A</td>
<td>Centrifugal pumps have significantly higher estimated flow accuracy.</td>
</tr>
<tr>
<td>Inlet suction</td>
<td>A &gt; C</td>
<td>Centrifugal pumps have significantly lower inlet suction at low flow conditions.</td>
</tr>
<tr>
<td>Ability to scale down</td>
<td>A &gt; C</td>
<td>Axial pumps can be more easily scaled down to sizes sufficient to be implanted intravascularly.</td>
</tr>
<tr>
<td>Pre-load sensitivity</td>
<td>A = C</td>
<td>Axial and centrifugal continuous-flow pumps both have low preload sensitivity relative to the native ventricle and pulsatile VAD.</td>
</tr>
<tr>
<td>After-load sensitivity</td>
<td>C &gt; A</td>
<td>Axial and centrifugal continuous flow pumps both have high after-load sensitivity relative to the native ventricle and pulsatile VAD; however, centrifugal pumps, by hydraulic performance characteristics, have higher after-load sensitivity.</td>
</tr>
<tr>
<td>Susceptibility to infection</td>
<td>A = C</td>
<td>No difference.</td>
</tr>
<tr>
<td>Biocompatibility</td>
<td>A = C</td>
<td>No difference.</td>
</tr>
<tr>
<td>Hemolysis</td>
<td>A = C</td>
<td>Not enough clinical data to suggest one is superior over the other.</td>
</tr>
<tr>
<td>Anti-coagulation</td>
<td>A = C</td>
<td>Not enough clinical data to suggest one is superior over the other.</td>
</tr>
<tr>
<td>Ability to wean the pump</td>
<td>A = C</td>
<td>Not enough clinical data to suggest one is superior over the other.</td>
</tr>
</tbody>
</table>

A, axial; C, centrifugal; VAD, ventricular assist device.

References


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