Engineering and Clinical Considerations in Rotary Blood Pumps

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In this chapter, general considerations and the operation principle of rotary blood pumps will be first presented with particular focus on the pressure-flow-speed characteristics, on what influences the pump flow rate, and on biocompatibility aspects. Finally current state-of-the-art about hemodynamic monitoring and control of these pumps will be presented.

14.1 General Considerations

Rotary blood pumps are used in the treatment of heart failure. Common indication for the implantation of these devices is end-stage heart failure (NYHA class IV) with expected 50% mortality within 1 year. Typically patients present with a low left ventricular ejection fraction (<25%), elevated pulmonary pressures, reduced cardiac index (<2 l/min/m^2), peak-VO_2 <12–14 ml/min/kg, chronic or intermitted inotropic dependence, and secondary progressive hepatic or renal dysfunction [1]. Rotary blood pumps can be used to bridge the patient until heart transplantation becomes possible, or they can even be implanted for lifetime. This latter is described as destination therapy and is considered when there is a contraindication for cardiac transplantation, such as irreversible pulmonary hypertension, active systemic infection, active malignancy or history of malignancy with probability of recurrence, or inability to comply with complex medical regimen. In a few cases, these devices can be used as bridges to recovery, such as in case of acute cardiac failure following cardiac surgery or acute myocarditis infections.

14.2 Working Principle and Classification

Rotary blood pumps are electromagnetically actuated turbodynamic machines [2]. These pumps consist of two main parts: a rotating component (impeller, i.e., a disk/annulus with vanes) and a pump housing. By its rotation, the impeller transfers energy from the electric motor that drives the pump to the blood. As a result of the impeller action, the blood leaves the impeller at a higher pressure and velocity than at its entrance. The impeller is supported within the pump housing by a bearing. In addition, a rotary blood pump assembly comprises inflow and outflow cannulas for its connection to the cardiovascular system and a flexible driveline for connection to electric power supply and to a control unit (Fig. 14.1).

Rotary blood pumps can be classified according to five main factors: geometry, bearing type, implantability, intended duration of use, and intended support function. Concerning the design geometry, one can distinguish three different types according to the flow path through the pump. If the angle between blood inflow and blood outflow is 90° (blood exits the pump in a direction orthogonal to the blood inflow), one speaks of a centrifugal-flow pump. If this angle is 0° (blood enters and leaves the pump along the same axis), one has an axial-flow pump. A pump characterized by angles between these two extreme cases is called a mixed-flow pump. There are three main types of bearing used to support the impeller of a rotary blood pump: mechanical, magnetic, and hydrodynamic bearings. The first bearing type relies on the low friction coefficient of the bearing material (ceramic, ruby), the second on magnetic forces, and the third on hydrodynamic forces to obtain levitation of the impeller and contactless rotation. Concerning implantability, one can distinguish between implantable devices, where the pump housing and cannulas are placed into the body with power supply and driving unit being still extracorporeal, and external devices, where the only implantable components are the pump inflow and outflow cannulas. The duration of use constitutes another factor to distinguish devices: one can have short-term devices that are intended for days or weeks and long-term devices for months or years of implantation. Finally, one can distinguish pumps that support the function of one ventricle (left or right ventricular assist devices, LVAD, RVAD), of both ventricles (BiVAD), or for heart replacement (rotary total artificial heart, rTAH). The focus in the following text will be on rotary pumps used as implantable LVADs, since these are the most commonly developed and used devices.

An implantable LVAD has its inlet cannula typically placed within the left ventricle, but cannulation to the left atrium is sometimes also used. The outlet cannula is commonly sutured to the ascending aorta, but the descending aorta or subclavian cannulation is also used. In Fig. 14.1, an implantable, axial-flow LVAD with mechanical

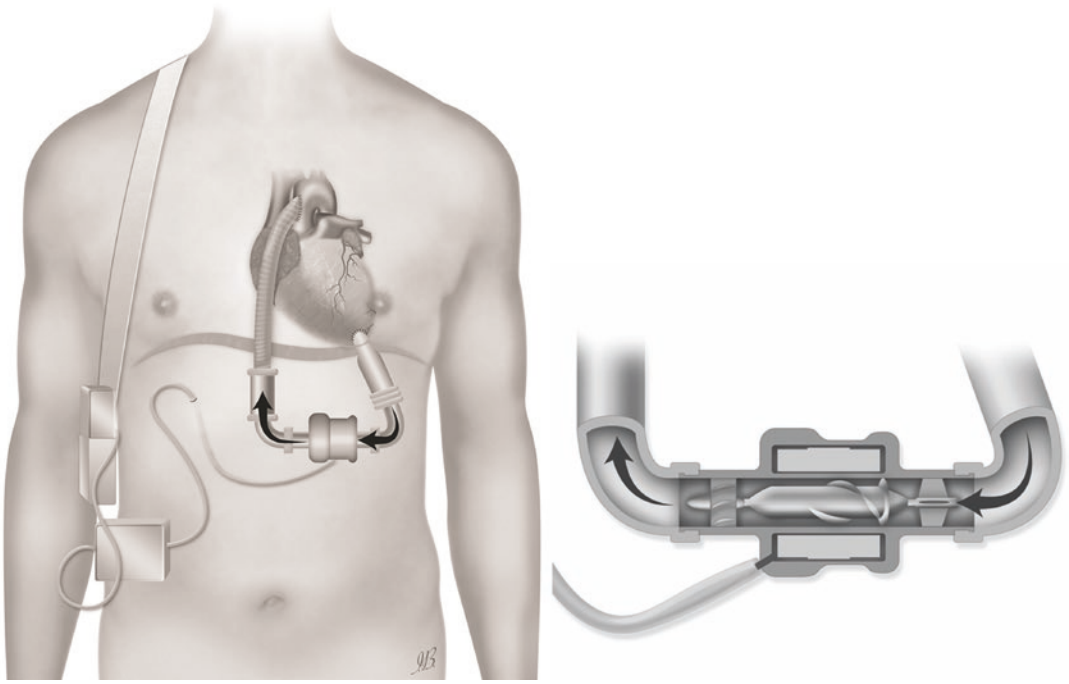

Fig. 14.1 LVAD implanted between the left ventricle and the ascending aorta (*left*); schematic diagram of the different components of an axial-flow pump (*right*) (Illustration by Ilaria Bondi's Peppermint Advertising)

bearings for long-term use is shown. The inflow is cannulated to the left ventricle and outflow to the ascending aorta.

14.3 Functional Requirements

The key functional requirements for rotary blood pumps can be summarized in the following bullet points: They should:

- Generate enough blood flow rate, 5–8 l/min, at physiological arterial pressure (100–150 mmHg for LVADs).
- Adapt to patient hemodynamic requirements (at least have a Starling-like behavior).
- Prevent hemodynamic overload to the right ventricle (in case of LVADs).
- Guarantee right/left flow balance (in case of BiVADs or rTAH).
- Be anatomically compatible with the large variations in patient size (body mass, chest diameter, abdominal girth).
- Be small to reduce surgical trauma and chronic infections at implantation site.
- Be structurally stable in corrosive environments.
- Guarantee continuous operation without maintenance for years (5–10 years).
- Minimize red blood cell damage (hemolysis) and activation of the coagulation cascade.
- Avoid formation of thrombi (all current devices require anticoagulation therapy).
- Be efficient for low power consumption and prolonged battery life.
- Have user-friendly interfaces and deliver unambiguous warnings and alarms.
- Be cost-effective (in terms of device, implantation, usage costs).

Modern rotary blood pumps fulfill most of the abovementioned requirements. They are all powered by a so-called brushless DC motor and have an external power supply: typically two rechargeable batteries or a power cord. The efficiency of these devices allows power consumption up to 10 W and a battery capacity that allows several hours of untethered patient activity. Rotary blood pumps are small in size, which allows minimally invasive implantation. They have with just one moving part (the impeller) and no valves as in the pulsatile counterpart, which increase the durability. This also allows silent

operation, which is important for patient quality of life. They suffer however from a lack of adaptation to changing hemodynamics, which leads, for example, to pump flow rate decrease in response to an increasing arterial pressure or to suck-down of the ventricular structures into the inflow cannula in responsed to a decreasing ventricular filling. They require therefore careful monitoring. Thrombus formation, strokes and bleeding still remain an issue with these devices, which although optimized for blood pumping are still challenging for coagulation and hemostasis [3, 4]. Further complications include infections of the percutaneous driveline.

14.4 Pressure-Flow-Speed Characteristics

In the following section, the typical operation of a rotary blood pump used as an LVAD will be presented. Circulatory support by a LVAD depends on the interaction between the residual ventricular function, the overall hemodynamics and the pump speed setting. Generally one distinguishes between partial support and full support. In partial support, the LVAD and the ventricle both eject blood toward the aorta (see Fig. 14.2, left panel). In case of full support, the LVAD alone pumps the whole cardiac output toward the aorta, and the aortic valve stays permanently closed (see Fig. 14.2, right panel). In both support types, the flow rate generated by the rotary pump is related to the ventricular and aortic pressures as well as to the pump speed.

In order to understand the interaction between the LVAD and the assisted heart, some definitions will be given first, and then the pressure-flow-speed characteristics will be introduced. The volume of blood pumped in one min by the rotary pump is referred to as the pump capacity or flow rate or even simply named pump flow. It is usually symbolized by the letter

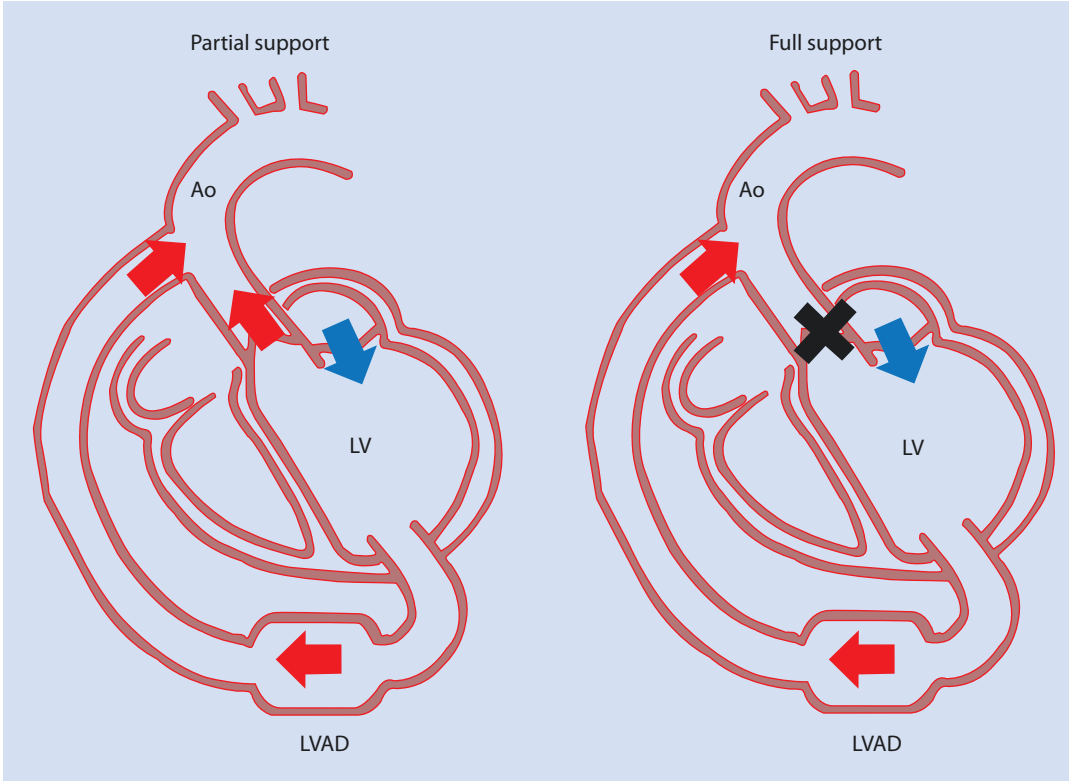

Fig. 14.2 Partial and full support by a left ventricular assist device (*LVAD*) in ventriculo-aortic cannulation. In case of partial support, the left ventricle (*LV*) and the LVAD pump blood in parallel toward the aorta (*Ao*). In case of full support, the LVAD alone pumps the whole cardiac output toward the Ao, and the aortic valve stays permanently closed (symbolized by *black cross*)

Q, and it is measured in liters per minute (l/min). The pressure difference between pump outlet and inlet is referred to as the head or simply pressure difference (outlet pressure=aortic pressure AoP, inled pressure=left ventricular pressure LVP). This difference is symbolized by the letter *H* or sometimes by ΔP, and it is usually measured in millimeters of mercury (mmHg). The speed at which the impeller rotates is referred to as pump speed. It is symbolized by the letter *N*, and it is measured in revolutions per minute (rpm). A rotary blood pump can be uniquely described by a relationship between these three variables. Specifically, variation of the pressure difference with pump flow at constant speed is called the pump characteristic. Considering the different speeds, one speaks of the pressure-flow-speed characteristics. In Fig. 14.3, a schematic diagram depicting characteristics for a hypothetical centrifugal-flow pump is shown. For a more technical description of hydraulic characteristics and design concepts of centrifugal- and axial-flow pumps, please refer to [2].

14.5 Pump Flow Rate-Influencing Factors

Pump characteristics are extremely useful to understand how much blood the assist device will pump at a given rotational speed and pressure difference/head between outlet and inlet (H = AoP – LVP). In this paragraph, a simplified and graphical analysis of the influence of these factors on the pump flow rate will be presented.

Due to a residual ventricular contractile function, the LVP (and to some extent the AoP) will pulsate during the cardiac cycle, leading to a periodic change of the pump pressure head (Hs and Hd in Fig. 14.4a). This pulsating pressure head will lead, via the pump characteristic for a given rotational speed, to a pulsatile pattern of the pump flow rate (Qs and Qd in Fig. 14.4b, c).

A rotary blood pump provides a continuous unloading of the ventricle since blood is continuously pumped through the heart cycle (the flow rate is >0). The blood flow rate retains a certain pulsatility that depends on the residual

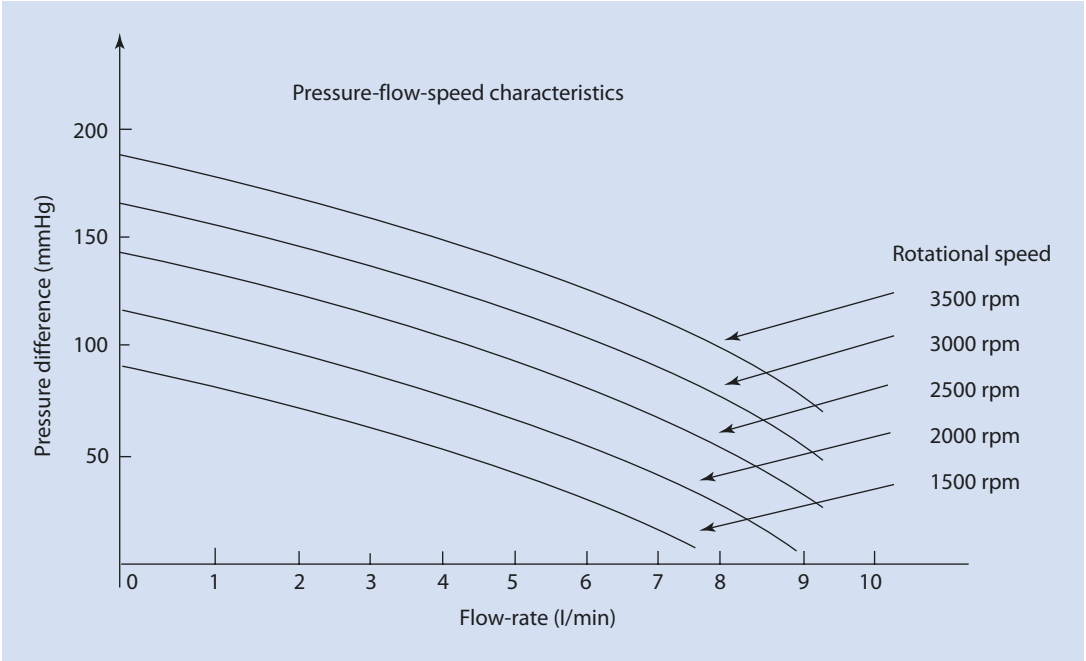

Fig. 14.3 Schematic drawing of pump characteristics (*solid lines*) of a hypothetical centrifugal-flow pump. For an LVAD connected between the left ventricle and the aorta, the pressure difference across the pump is equal to the aortic pressure minus the left ventricular pressure

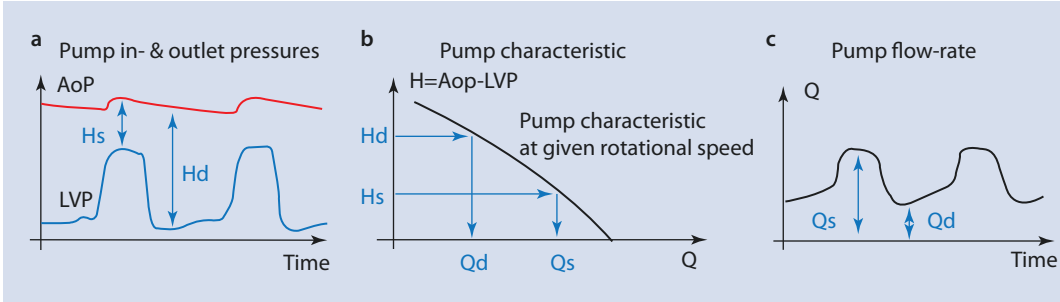

Fig. 14.4 In the leftmost panel (**a**), the time course of pressures at the LVAD outlet and inlet as well as their difference during systole (*Hs*) and diastole (*Hd*) is shown. In the middle panel (**b**), a pump characteristic relating the Hs and Hd to the systolic and diastolic flow rates (*Qs* and *Qd*) is shown. In the rightmost panel (**c**), the time course of the LVAD flow rate is shown

contractile function of the assisted ventricle. It is therefore incorrect to name rotary blood pumps as "nonpulsatile" assist devices. Only in the rare case of ventricular fibrillation (no residual contraction) these devices will deliver a truly nonpulsatile flow rate. It must be however noted that the flow rate pulsatility that these devices are able to provide is often small compared to native pulsatility. This reduced flow pulsatility leads to small arterial pressure pulse, which can be difficult to measure using conventional auscultation or automated cuffs. Instead Doppler ultrasound can be used to detect flow in the radial artery when the cuff pressure becomes lower that the systolic arterial pressure [5].

By means of the simple three diagrams presented in Fig. 14.4, the effect of changes in pump speed, pump preload (LVP), and pump afterload (AoP) can be analyzed. In the following analysis, a condition of full support (see Fig. 14.2, right panel) is considered. An increase in the three abovementioned variables is considered here; when a decrease occurs, the opposite changes will take place.

In Fig. 14.5a, the effect of a pump speed increase at constant pre- and afterload is shown. Because of constant pre- and afterload, the systolic and diastolic pressure head are constant too. The new pump characteristic at a higher speed will lead therefore to an increase in both systolic and diastolic flow rates (Qd' and Qs' in Fig. 14.5a – middle and rightmost panels). This increase in flow rate will lead to circulatory adaptation and consequent later changes in LVP and AoP (mostly by reducing LVP and increasing AoP). The assumption of constant preload and afterload is therefore strictly valid in the short time after the speed change; the later changes can be investigated in a similar manner, however, as it is presented next.

In Fig. 14.5b, the effect of an increased pump preload (LVP) at constant afterload and speed is depicted. Pump preload can increase, for example, due to increased venous return to the LV or increased LV contractility. With a change in venous return or contractility, the systolic LVP will typically increase much more than the diastolic one, leading to a reduction of the Hs and an almost constant Hd (Fig. 14.5b leftmost panel). This new hemodynamic conditions will lead to an increased systolic flow rate (Qs' in Fig. 14.5b middle and rightmost panels) and consequently an increased waveform pulsatility (defined as the difference between systolic and diastolic flow rates Qpuls=Qs-Qd). In this analysis, the pump afterload is considered constant. Also this assumption is strictly valid in the short time after the preload change, and a further analysis can be performed considering afterload effects presented next.

Finally, in Fig. 14.5c the effect of an increased afterload (AoP) at constant preload and speed is shown. The increase of AoP leads to an increase of both Hs and Hd (Fig. 14.5c leftmost panel), that leads, via the pump characteristics, to a decrease in pump flow (Qs' and Qd' Fig. 14.5c middle and rightmost panels). This decrease in pump flow will lead a change in the overall hemodynamics, possibly leading to insufficient ventricular unloading and blood accumulation in the pulmonary circulation.

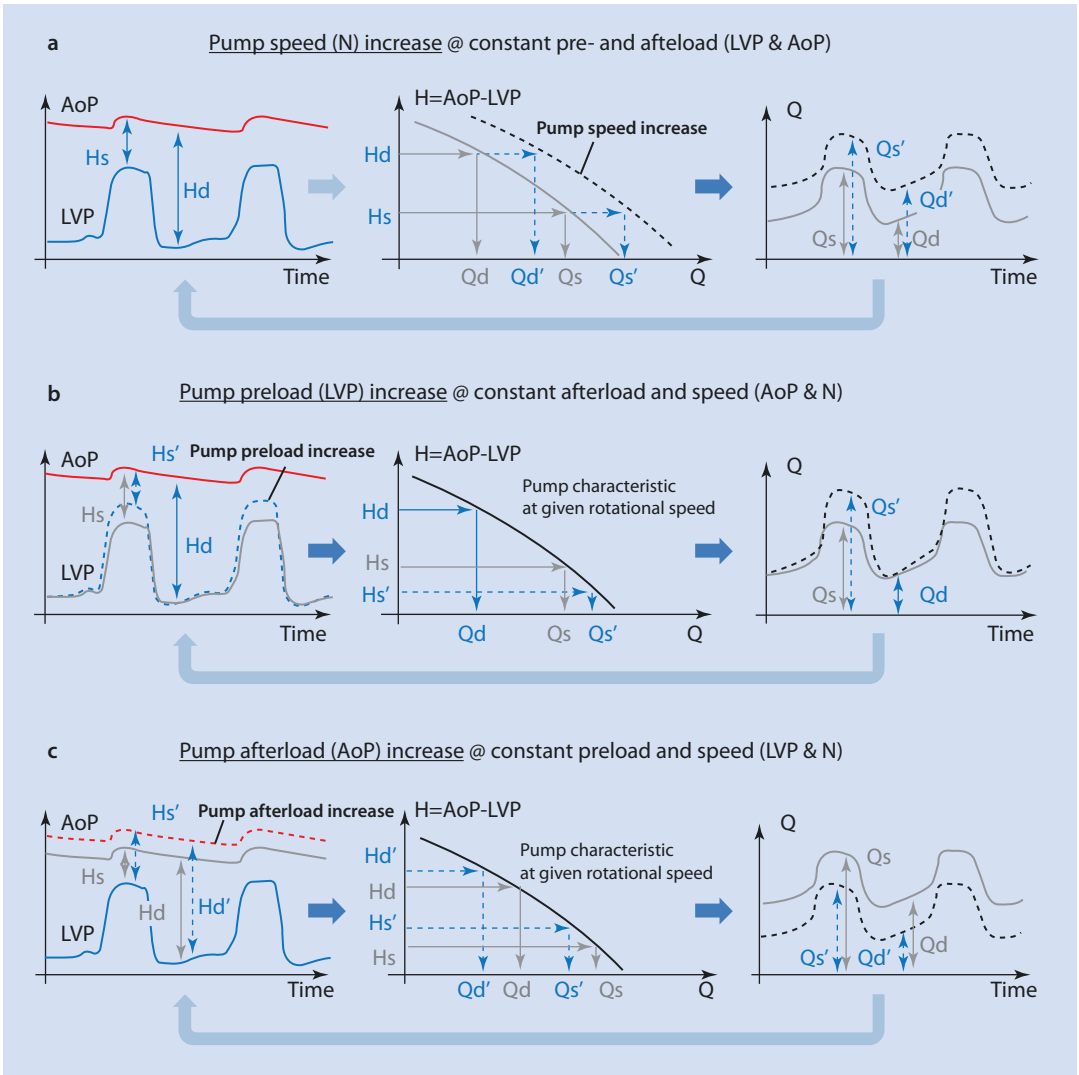

Fig. 14.5 Effect of isolated changes in speed (*panel a*), preload (*panel b*), and afterload (*panel c*) on the pump flow rate. The arrows feeding back from the rightmost to the leftmost panel mean that a change in pump flow will result in a change in the overall hemodynamics, thus a change in LVP and AoP

The condition of partial support was not considered above. The analysis is however similar with just one relevant difference. When the aortic valve opens, the systolic pressure head (AoP-LVP during systole) will be approximately zero (slightly negative), and it remains rather independent on pre- and afterload changes. In this case, the systolic pump flow rate will have a rather constant value that depends only on the speed setting. The systolic flow rate will be indeed the intercept of the pump characteristic with the x-axis (e.g., in Fig. 14.3, $Qs = 9$ l/min for $N = 2000$ rpm).

14.6 Blood Trauma and Thrombogenicity

The probably most crucial aspect in the development of rotary blood pumps is the blood compatibility, both concerning mechanical blood trauma and triggering platelet activation and coagulation of thrombi.

In the early developments, spinning disks were preferably used to avoid high shear stress exposure [6]. It is known since long time that the damage of erythrocytes depends on shear stress

value, exposure time, and type of flow [7]. Therefore, an ultrashort exposure to high shear can be preferable to longer exposures to less shear [8], which explains the shape of many of the impellers of rotary pumps, which have different (e.g., straight) vane geometries compared to classic rotary pumps [9, 10] . However, the trauma (far less than one erythrocyte out of 1 mio should be destroyed per passage) is difficult to measure, and short exposure times at well-defined shear would require new test setups, which are not available yet. As a consequence, even after decades of research, numerical models of blood trauma are limited in reliability and accuracy [11].

Thrombogenicity and particularly platelet activation can be caused by local shear stress, hot spots due to bearing/sealing friction, surface roughness, and long residence times in stagnation areas [12]. To provide thrombogenicity tests for pumps and specific blood pathways, several in vitro setups have been developed [13, 14]. Mechanical bearings have been identified as the most critical parts in design, due to their shear gradient, the generated friction heat, and critical location in the center of the rotor in usually low-flow areas. As a potential solution to this issue, pumps with either actively controlled magnetic bearings or combinations of permanent magnets with hydrodynamic bearings have been developed and successfully implemented in commercially available devices.

14.7 Hemodynamic Monitoring and Physiological Control

14.7.1 LVAD Patient Monitoring

A detailed knowledge of the hemodynamic interaction of the pump and the cardiovascular system allows one to predict hemodynamic behavior depending on the pump flow rate waveform and therefore perform patient monitoring using pump data. This is especially important in rotary blood pumps because of their lack of adaptation to hemodynamic changes. Apart from standard clinical diagnostics [5], approaches to monitor the LVAD patient can be divided in those which make use of external sensors and those that rely on assist device motor parameters to estimate hemodynamic variables [15]. Monitoring using an implantable pump flow rate sensor has been reported in patients [16]. Developments of pressure sensors [17, 18] as well as impedance sensors [19] to be embedded in LVAD systems have been reported. In [20] the possibility of utilizing in LVAD patients remote pressure monitoring tools designed for non-LVAD heart failure patients is reviewed. Patient monitoring that relies only on available pump data seems very promising, particularly because additional sensors are often affected by drifts and may be prone to failure. Several hemodynamic indices and methods were developed based on available pump signals: a ventricular contractility index [21], a ventricular relaxation index [22], as well as the discrimination between full and partial assistance (state of the aortic valve opening during support) [23–26]. A method to evaluate heart rate and arrhythmic events (e.g., sustained ventricular tachycardia or atrial fibrillation) as well as heart rate variability [27] and methods to detect conditions of overpumping (i.e., ventricular suction) [28, 29] were also developed. All these methods and indices use either pump flow rate waveform, which can be either measured or estimated from pump motor current and speed [30], or directly the motor current/speed waveforms. In ◘ Fig. 14.6a, graphical overview of hemodynamic features that can be derived from the pump flow rate waveform is shown. These methods take advantage of the pulsatility in the pump flow rate that depends on the residual contractile function of the assisted ventricle (◘ Fig. 14.4).

Pump Physiological Control

Since the first days of clinical application, rotary pumps have been driven at constant speed. In this setting, the pump can be understood as a “turbo discharger” of the ventricle, which – simply speaking – facilitates ventricular output by providing pressure work. If the ventricle has some remaining contractile force or recovers after implant, it can overtake some physiological adaptation by the Starling mechanism still working against a lower output pressure. This explains why most patients can perform their daily activities and some reduced exercise at such constant speed. Only in patients with completely dysfunctional ventricle such adaptation is difficult, with circadian hemodynamic changes that may lead to suction during night and

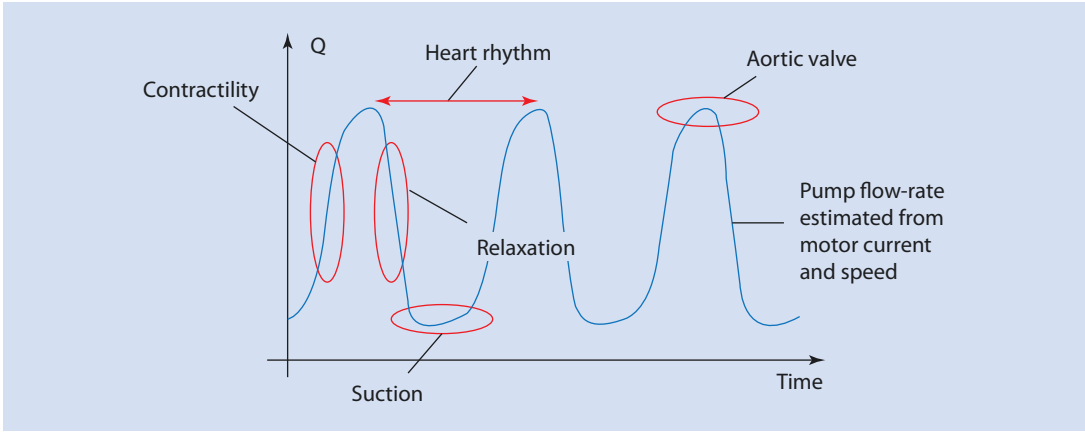

Fig. 14.6 Hemodynamic features that can be derived from an estimation of the pump flow rate waveform

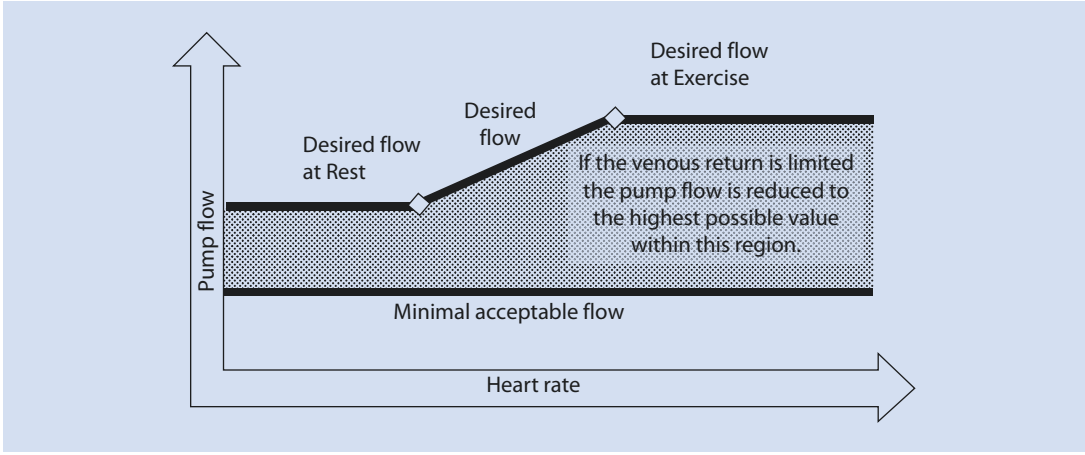

Fig. 14.7 Physiological controller settings and function. A target desired flow depending on heart rate is set by the physician. This is achieved by automatically adjusting pump speed. In case of reduced venous return (suction detected or too low flow pulsatility), the highest possible flow is automatically achieved. If this falls below a minimal acceptable level, set by the physician, a fail-safe mode is activated to avoid further decrease

insufficient supply during the day even at rest [31]. Especially for such weak patients, for early postoperative recovery, and particularly for activity and exercise, a physiologically adaptive speed control would be extremely desirable. The key problem for such algorithms however still is the lack of reliable, drift-free, durable, and biocompatible pressure transducers, which would allow a control based on inflow preload pressure. A control should also fulfil several targets: it should maintain atrial pressure within a physiological range, but at the same time avoid excessive ventricular unloading or suction and provide an overall aortic flow correlated to the demand as in a healthy individual. If a physiological control should allow also intermittent opening of the aortic valve, support pulsatility of aortic pressure, and provide myocardial protection or even myocardial training for recovery, is currently under debate.

To fulfil these requirements, many algorithms have been described and tested in silico and in vitro, but only very few made it into in vivo or even clinical tests. For a review of physiological control strategies for rotary LVADs, please refer to [32, 33]. A physiological control including suction detection and speed based on remaining pulsatility and heart rate has been successfully clinically tested. In Fig. 14.7, a schematic diagram of the controller settings and function is shown. For details about the controller design and the clinical study, please refer to [34, 35].

Although this control algorithm has proven stable in rest, exercise, Valsalva maneuvers, and even during severe arrhythmia, it has not been implemented in commercially available devices yet. Obstacles for an implementation into clinical routine are probably liability challenges and correlated questions of continuous recording of pump control activity ("how to prove the innocence of the control algorithm in case of problems?") and perhaps also the necessity for higher pump flow rates during exercise than current devices can deliver. Currently physicians and patients are satistifed with the basic support provided by LVADs, but this may soon change with increasing patient needs for better life quality related to full participation in life, physical activity, and improved usability of ventricular assist systems.

References

1. Miller LW, Guglin M (2013) Patient selection for ventricular assist devices: a moving target. J Am Coll Cardiol 61(12):1209–1221
2. Stepanoff AJ (1957) Centrifugal and axial flow pumps: theory, design, and application, 2nd edn. John Wiley & Sons, New York, p 462
3. Kirklin JK, Naftel DC, Kormos RL, Stevenson LW, Pagani FD, Miller MA et al (2013) Fifth INTERMACS annual report: risk factor analysis from more than 6,000 mechanical circulatory support patients. J Heart Lung Transplant Off Publ Int Soc Heart Transplant 32(2):141–156
4. Kirklin JK, Naftel DC, Pagani FD, Kormos RL, Stevenson LW, Blume ED et al (2014) Sixth INTERMACS annual report: a 10,000-patient database. J Heart Lung Transplant Off Publ Int Soc Heart Transplant 33(6):555–564
5. Estep JD, Trachtenberg BH, Loza LP, Bruckner BA (2015) Continuous flow left ventricular assist devices: shared care goals of monitoring and treating patients. Methodist Debakey Cardiovasc J 11(1):33–44
6. Bernstein EF, Dorman FD, Blackshear PL, Scott DR (1970) An efficient, compact blood pump for assisted circulation. Surgery 68(1):105–115
7. Schima H, Trubel, W, Muller, MR, Papantonis, D, Salat, A, Schlusche, A et al (1991) Development of a centrifugal blood pump with minimal hemolysis. In: Proceedings of the International Workshop on Rotary Blood Pumps
8. Wampler RK, Moise JC, Frazier OH, Olsen DB (1988) In vivo evaluation of a peripheral vascular access axial flow blood pump. ASAIO Trans Am Soc Artif Intern Organs 34(3):450–454
9. Schima H, Müller MR, Papantonis D, Schlusche C, Huber L, Schmidt C et al (1993) Minimization of hemolysis in centrifugal blood pumps: influence of different geometries. Int J Artif Organs 16(7):521–529
10. Larose JA, Tamez D, Ashenuga M, Reyes C (2010) Design concepts and principle of operation of the HeartWare ventricular assist system. ASAIO J Am Soc Artif Intern Organs 56(4):285–289
11. Hariharan P, D'Souza G, Horner M, Malinauskas RA, Myers MR (2015) Verification benchmarks to assess the implementation of computational fluid dynamics based hemolysis prediction models. J Biomech Eng 137(9): 094501 (10 Pages)
12. Chiu W-C, Girdhar G, Xenos M, Alemu Y, Soares JS, Einav S et al (2014) Thromboresistance comparison of the HeartMate II ventricular assist device with the device thrombogenicity emulation- optimized HeartAssist 5 VAD. J Biomech Eng 136(2):021014
13. Schima H, Siegl H, Mohammad SF, Huber L, Müller MR, Losert U et al (1993) In vitro investigation of thrombogenesis in rotary blood pumps. Artif Organs 17(7):605–608
14. Dimasi A, Rasponi M, Sheriff J, Chiu W-C, Bluestein D, Tran PL et al (2015) Microfluidic emulation of mechanical circulatory support device shear-mediated platelet activation. Biomed Microdevices 17(6):117
15. Bertram CD (2005) Measurement for implantable rotary blood pumps. Physiol Meas 26(4):R99–117
16. Pektok E, Demirozu ZT, Arat N, Yildiz O, Oklu E, Eker D et al (2013) Remote monitoring of left ventricular assist device parameters after HeartAssist-5 implantation. Artif Organs 37(9):820–825
17. Bullister E, Reich S, D'Entremont P, Silverman N, Sluetz J (2001) A blood pressure sensor for long-term implantation. Artif Organs 25(5):376–379
18. Zhou M-D, Yang C, Liu Z, Cysyk JP, Zheng S-Y (2012) An implantable Fabry-Pérot pressure sensor fabricated on left ventricular assist device for heart failure. Biomed Microdevices 14(1):235–245
19. Her K, Ahn CB, Park SM, Choi SW (2015) Heart monitoring using left ventricle impedance and ventricular electrocardiography in left ventricular assist device patients. Biomed Eng Online 14:25
20. Lampert BC, Emani S (2015) Remote hemodynamic monitoring for ambulatory left ventricular assist device patients. J Thorac Dis 7(12):2165–2171
21. Naiyanetr P, Moscato F, Vollkron M, Zimpfer D, Wieselthaler G, Schima H (2010) Continuous assessment of cardiac function during rotary blood pump support: a contractility index derived from pump flow. J Heart Lung Transplant 29(1):37–44
22. Moscato F, Granegger M, Naiyanetr P, Wieselthaler G, Schima H (2012) Evaluation of left ventricular relaxation in rotary blood pump recipients using the pump flow aveform: a simulation study. Artif Organs 36(5):470–478
23. Bishop CJ, Mason NO, Kfoury AG, Lux R, Stoker S, Horton K et al (2010) A novel non-invasive method to assess aortic valve opening in HeartMate II left ventricular assist device patients using a modified Karhunen-Loève transformation. J Heart Lung Transplant Off Publ Int Soc Heart Transplant 29(1):27–31
24. Granegger M, Schima H, Zimpfer D, Moscato F (2014) Assessment of aortic valve opening during rotary blood pump support using pump signals. Artif Organs 38(4):290–297

25. Hayward C, Lim CP, Schima H, Macdonald P, Moscato F, Muthiah K et al (2015) Pump speed waveform analysis to detect aortic valve opening in patients on ventricular assist device support. Artif Organs 39(8):704–709
26. Granegger M, Masetti M, Laohasurayodhin R, Schloglhofer T, Zimpfer D, Schima H et al (2016) Continuous monitoring of aortic valve opening in rotary blood pump patients. IEEE Trans Biomed Eng 63:1201–1207
27. Moscato F, Granegger M, Edelmayer M, Zimpfer D, Schima H (2014) Continuous monitoring of cardiac rhythms in left ventricular assist device patients. Artif Organs 38(3):191–198
28. Vollkron M, Schima H, Huber L, Benkowski R, Morello G, Wieselthaler G (2004) Development of a suction detection system for axial blood pumps. Artif Organs 28(8):709–716
29. Vollkron M, Schima H, Huber L, Benkowski R, Morello G, Wieselthaler G (2006) Advanced suction detection for an axial flow pump. Artif Organs 30(9):665–670
30. Granegger M, Moscato F, Casas F, Wieselthaler G, Schima H (2012) Development of a pump flow estimator for rotary blood pumps to enhance monitoring of ventricular function. Artif Organs 36(8):691–699
31. Wang Y, Koenig SC, Slaughter MS, Giridharan GA (2015) Rotary blood pump control strategy for preventing left ventricular suction. ASAIO J Am Soc Artif Intern Organs 61(1):21–30
32. AlOmari A-HH, Savkin AV, Stevens M, Mason DG, Timms DL, Salamonsen RF et al (2013) Developments in control systems for rotary left ventricular assist devices for heart failure patients: a review. Physiol Meas 34(1):R1–27
33. Pauls JP, Stevens MC, Bartnikowski N, Fraser JF, Gregory SD, Tansley G (2016) Evaluation of physiological control systems for rotary left ventricular assist devices: an in-vitro study. Ann Biomed Eng 44:2377–2387
34. Vollkron M, Schima H, Huber L, Benkowski R, Morello G, Wieselthaler G (2005) Development of a reliable automatic speed control system for rotary blood pumps. J Heart Lung Transplant Off Publ Int Soc Heart Transplant 24(11):1878–1885
35. Schima H, Vollkron M, Jantsch U, Crevenna R, Roethy W, Benkowski R et al (2006) First clinical experience with an automatic control system for rotary blood pumps during ergometry and right-heart catheterization. J Heart Lung Transplant Off Publ Int Soc Heart Transplant. 25(2):167–173