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STUDY OF THE HEAD-CAPACITY CURVES OF AN AXIAL PUMP UNDER HIGH BACK-PRESSURE CONDITIONS

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Objective: to study the hydraulic characteristics of an axial pump under conditions where the external back pressure at the outlet exceeds the maximum pressure generated by the pump, and to determine the corresponding head-capacity curves (HCC) in this operating mode. **Materials and methods.** A hydraulic test bench was designed and assembled for the experiments, capable of generating outlet back pressure up to 200 mmHg. Measurements were performed using axial blood pump Stream Cardio (Russia). A series of tests was conducted at rotational speeds ranging from 7,500 to 10,000 rpm. Initial excess pressures of 85, 100, and 120 mmHg were applied. **Results.** The experimental data obtained on the hydraulic test bench made it possible to determine the complete HCC of the axial pump, including the reverse-flow region corresponding to negative flow rates. At a flow rate of 0 L/min, the pressure drop across the pump was 85 mmHg. At a pressure drop of 100 mmHg, the flow rate reached –2.5 L/min, and at 120 mmHg it reached –5 L/min. The HCC in the negative flow range represents a continuous extension of the pump characteristic curve, preserving a similar slope and exhibiting an approximately linear behavior.

Keywords: heart failure, mechanical circulatory support, pump flow rate, pump pressure, axial pump, reverse flow.

INTRODUCTION

Implantable cardiac assist devices are widely used in clinical practice to manage end-stage chronic heart failure. They serve as a critical bridge to heart transplantation or as destination therapy in patients with contraindications to transplantation [1]. Several commercially available systems are currently in routine use worldwide. Among the most common systems are HeartMate III and HeartMate II (Thoratec Corp., USA), as well as HeartWare (Medtronic, USA) [2].

There is a growing demand for devices whose technical specifications enable prolonged mechanical circulatory support (MCS) while maintaining a high quality of life for patients. A key engineering parameter of such systems is the head-capacity curve (HCC), which describes the relationship between the pressure head generated by the device and the resulting flow rate. The shape of the HCC is largely determined by the design of the flow path within the pump. For example, Fig. 1 illustrates the HCC of the axial-flow HeartMate II pump and the centrifugal-flow HeartMate III pump within their typical operating ranges [3]. Axial-flow pumps are characterized by a flatter relationship, often approximated by a hyperbolic function. In contrast, centrifugal pumps, such as the HeartMate III, exhibit a logarithmic relationship ($y = \log_a x$, where $0 < a < 1$). Analysis of these specifications shows that, at the same flow rate of 3 L/min, the HeartMate II pump generates a pressure head of approximately 70 mmHg, whereas the HeartMate III produces approximately 85 mmHg. Thus, at the same

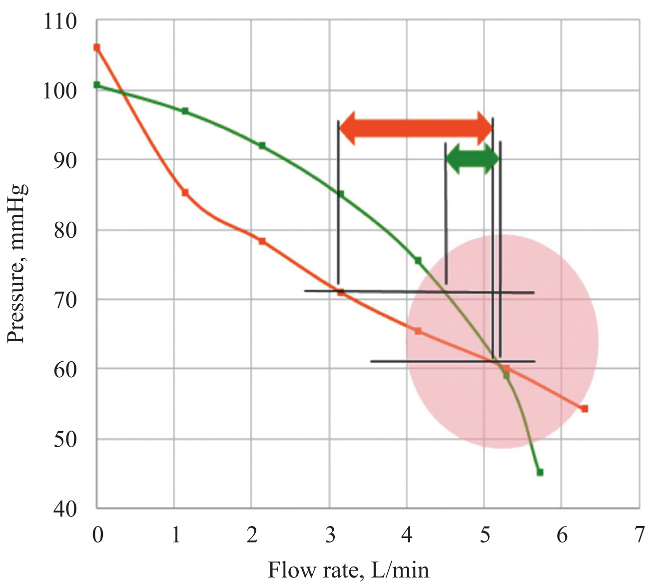

Fig. 1. Head-capacity curves of the HeartMate II (9,000 rpm; axial-flow pump, orange line) and HeartMate III (5,400 rpm; centrifugal-flow pump, green line). The orange circle indicates the typical operating range

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flow rate, the two systems operate at different pressure heads and, consequently, different outlet pressures, resulting in distinct hemodynamic effects within this range.

In the region highlighted in orange (Fig. 1), the flow–pressure characteristics of both pumps converge. At a flow rate of approximately 5 L/min, each system generates a pressure gradient of about 60 mmHg, indicating comparable hemodynamic performance. Under these conditions, aortic pressure pulsations occurring during systole, when the centrifugal HeartMate III pump operates in parallel with the native heart, exert a reduced influence on overall cardiac output compared with the axial-flow HeartMate II system.

Fig. 2 presents the HCCs of the left ventricular pump (LVP) integrated into the medical device "Universal System for Mechanical Support of the Pumping Function of the Left and Right Ventricles of the Heart" (Dona-M LLC, Russia). These characteristics are shown as a family of curves describing the relationship between the pressure head generated by the LVP and the flow rate at different rotor speeds.

Within the rotational speed range of 7,500–10,000 rpm, the pump generates a pressure head of approximately 45–85 mmHg at a flow rate of 5 L/min. A reduction in rotor speed below 7,500 rpm is associated with a corresponding decrease in the pressure head produced by the pump.

Most studies describing HCCs focus primarily on the positive flow range. However, under real-world conditions, the opposite scenario may occur during pump operation. In the presence of pulsatile flow generated by the left ventricle, transient increases in aortic pressure may exceed the pressure head produced by the pump. This can result in reverse (retrograde) flow through the device, a phenomenon that requires careful consideration and further investigation.

Such situations have been reported when systemic arterial pressure exceeds the pressure generated by the MCS system for various reasons [4–6]. This imbalance can typically be corrected by increasing the pump rotor speed, thereby restoring adequate hemodynamics. For example, in the case of the LVP, at a pressure difference of 100 mmHg and a rotor speed of 8500 rpm (Fig. 2), the flow rate is approximately 1 L/min. If the pressure exceeds 100 mmHg at the same rotor speed, the flow rate decreases further and may become negative, indicating the onset of reverse flow.

Several studies have reported that a reduction in pump rotor speed may lead to reverse blood flow through the device [7–10]. Ultrasound evaluation of this phenomenon in the HeartMate II pump demonstrated a bidirectional flow pattern (Fig. 3), characterized by a positive spectrum during systole and a negative spectrum during diastole [11].

Under physiological conditions, at the onset of diastole, aortic pressure remains relatively high, while left ventricular pressure is low, facilitating ventricular filling. The aortic valve prevents retrograde blood flow from the aorta into the left ventricle. However, during MCS, an alternative pathway for blood flow between the left ventricle and the aorta is established. Similar to the function of the left ventricle, the mechanical pumping system creates a direct flow of blood from the left ventricle to the aorta. However, the mechanical support system lacks a valve mechanism, which can lead to blood flowing in the opposite direction from the aorta back into the left ventricle.

Since the pump operates in parallel with the native heart, a physiological pressure difference may develop during diastole in which aortic pressure exceeds left ventricular pressure and may surpass the maximum pressure head generated by the pump. Under MCS conditions, the function of the aortic valve is indirectly maintained through appropriate regulation of pump operating parameters. Specifically, it is essential to ensure that the pressure generated at the pump outlet supports forward (positive) flow at a given flow rate.

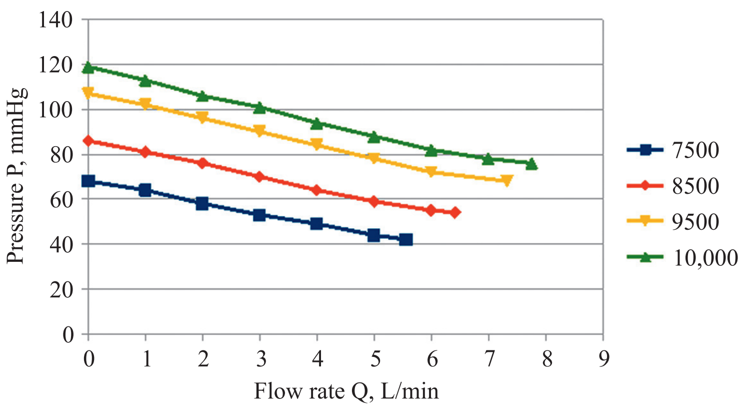

Fig. 2. Head-capacity curves of an axial pump at rotor speeds ranging from 7,500 to 10,000 rpm. A pronounced, nearly linear plateau is observed across all operating modes, indicating high sensitivity of pump blood flow to pressure fluctuations in the aorta

The required parameters are achieved by adjusting the rotor speed. The corresponding pressure and flow characteristics at different speeds can be derived from the pump's HCC or monitored via the device control system. However, reverse flow remains a complex challenge, requiring not only optimization of device settings but also appropriate medical management to stabilize systemic arterial pressure.

The objective of this study is to evaluate the performance of an axial-flow pump under conditions in which systemic pressure exceeds the pressure generated at the pump outlet, to identify the HCCs at which backflow is likely to occur, and to come up with recommendations for optimizing pump rotor speed.

MATERIALS AND METHODS

A dedicated experimental test bench (Fig. 4) was developed to evaluate pump performance under high outlet (back) pressure.

The system consisted of two tanks (tank 1 and tank 2) connected by a 1/2-inch diameter flexible tubing line. The axial-flow blood pump under investigation was installed within the main circuit. The direction of forward (positive) flow during pump operation is indicated by arrow 1 in the figure.

Flow measurements were performed using a dual ultrasonic flow sensing unit (Flowmeter; SONOFLOW CO.55/160, Germany) mounted on the main line. Because each sensor is capable of measuring flow in only one direction, two sensors were arranged in opposing orientations to enable detection of both forward and reverse (negative) flow.

Pressure difference between the pump inlet and outlet was recorded using a differential pressure sensor (Pressure Sensor; Motorola MPX5050, Japan). Tank 1 was equipped with a manual pneumatic pump and a sealed design, allowing controlled generation of elevated hydrostatic pressure within the system. Fluid pressure in

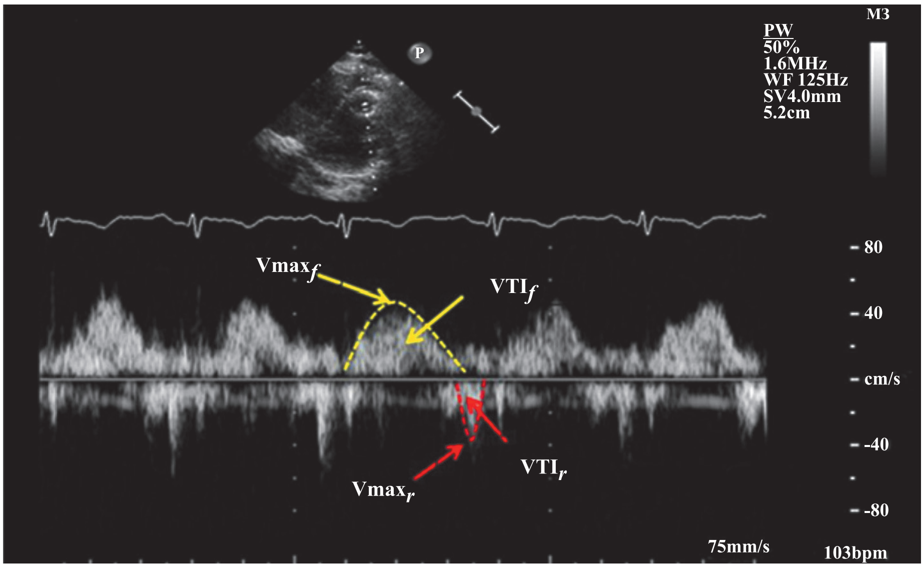

Fig. 3. Signal spectra for positive flow (yellow) and negative flow (red) regimes of the axial pump

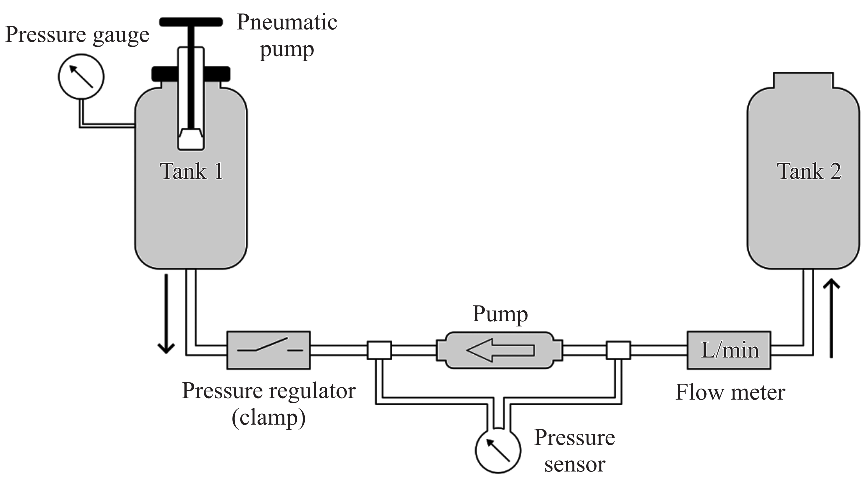

Fig. 4. Schematic diagram of hydraulic test bench

the system was monitored by measuring the air pressure above the liquid surface using a mechanical dial pressure gauge. Signals from the ultrasonic flow sensors and the differential pressure sensor, along with operational parameters of the pump – including average rotor winding current (reflecting the torque generated by the rotor), average rotor speed, and total current consumption of the pumping system – were recorded via an analog-to-digital converter module (E14-140, L-CARD). Data were acquired at a sampling frequency of 20 Hz and stored on a personal computer for subsequent processing and analysis using specialized software.

Tank 1 was filled with 10 L of distilled water to ensure complete filling of the system and removal of air from the circuit. The fluid-filled tubing and the axial pump were initially isolated using a pressure regulator positioned at the pump outlet. An initial overpressure was then generated inside tank 1 using a manual pneumatic pump. Following system preparation, the axial pump was initiated at a predefined rotor speed set by the control unit.

A series of experiments was conducted at rotor speeds ranging from 7,500 to 10,000 rpm. The initial gauge pressure varied between 85 and 200 mmHg.

RESULTS

The temporal profiles of backflow rate (Q, L/min) and backpressure (P, mmHg) at a rotor speed of 8,500 rpm are presented in Fig. 5.

The figure demonstrates four distinct phases of system behavior:

1. Phase 1 (0–25 s). Steady-state equilibrium prior to opening the main line. The circuit is closed, flow rate is zero, and pressure drop across the pump is 85 mmHg, which corresponds to the pump's HCC at a rotor speed of 8,500 rpm (Fig. 2).
2. Phase 2 (25–26 s). Transient response following opening of the main line. A rapid increase in outlet pressure is observed, accompanied by a sharp rise in backflow rate due to high backpressure. The flow rate enters the negative range, indicating reverse flow. This occurs as fluid from tank 1, where a pre-established elevated pressure of 200 mmHg exists, enters the pump inlet. This pressure significantly exceeds the 85-mmHg pressure drop predicted by the HCC at 8,500 rpm. Consequently, fluid is driven through the pump from the high-pressure region (tank 1) toward the low-pressure region (tank 2).
3. Phase 3 (26–35 s). Steady-state reverse flow. This phase is characterized by sustained retrograde flow under approximately constant operating conditions of pressure and flow rate, corresponding to the primary objective of the experiment. During this interval, fluid pressure in tank 1 gradually decreases due to reduction in fluid volume and the associated decrease in air pressure within the sealed chamber.
4. At this point, the pump develops a pressure head of 85 mmHg at a rotor speed of 8,500 rpm, consistent with its HCC. At this point, there is no fluid flow in either the forward or reverse directions (flow rate = 0). The pump thus operates in a state functionally analogous to a closed aortic valve at a systemic pressure of 85 mmHg. In this situation, a further decrease in systemic pressure results in forward flow into the aorta, whereas an increase above the pump-generated pressure head leads to reappearance of retrograde flow.

The data obtained during Phase 3 allow us to plot the pressure–flow P(Q) relationship and determine the HCC for the negative flow region. Fig. 6 presents the pressure drop generated by the pump (P) as a function of backflow

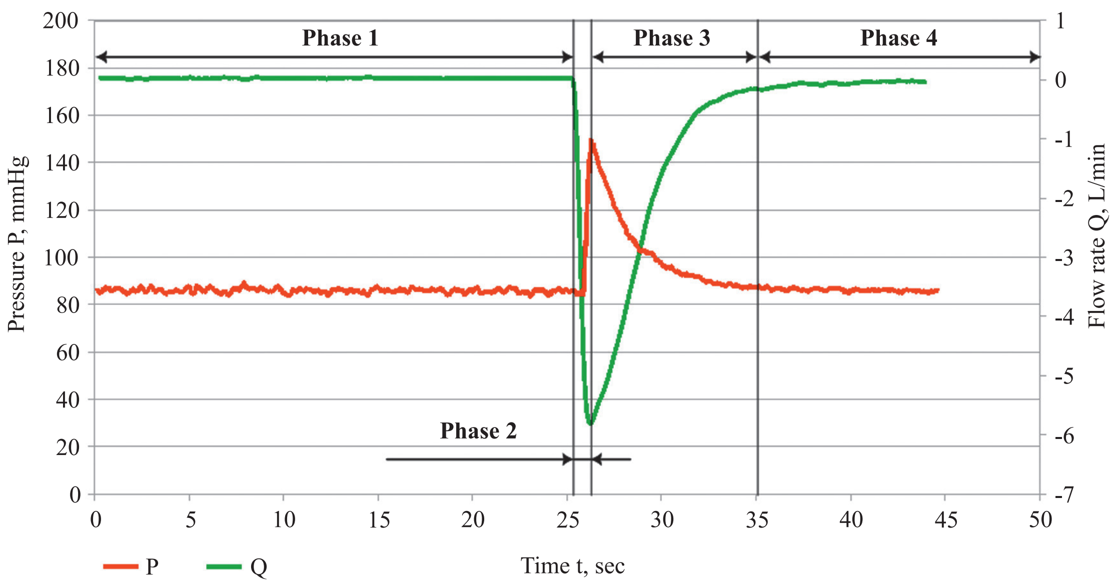

Fig. 5. Reverse flow rate through the axial pump under excess back pressure (rotational speed 8,500 rpm; initial back pressure 200 mmHg) after mathematical processing

rate (Q) at different initial pressures (85–200 mmHg in Tank 1) and a constant rotor speed of 8,500 rpm.

At zero flow (Q = 0 L/min), pressure drop across the pump is 85 mmHg. With increasing reverse flow, the system exhibits a progressive increase in pressure drop: at a pressure of 100 mmHg, flow rate is –2.5 L/min, and at 120 mmHg, it reaches –5 L/min. Further increases in backflow result in an approximately linear rise in pressure drop. As shown in Fig. 6, the initial pressure primarily shifts the range of measurable values, while the HCC values, which are close to linear, remain unchanged.

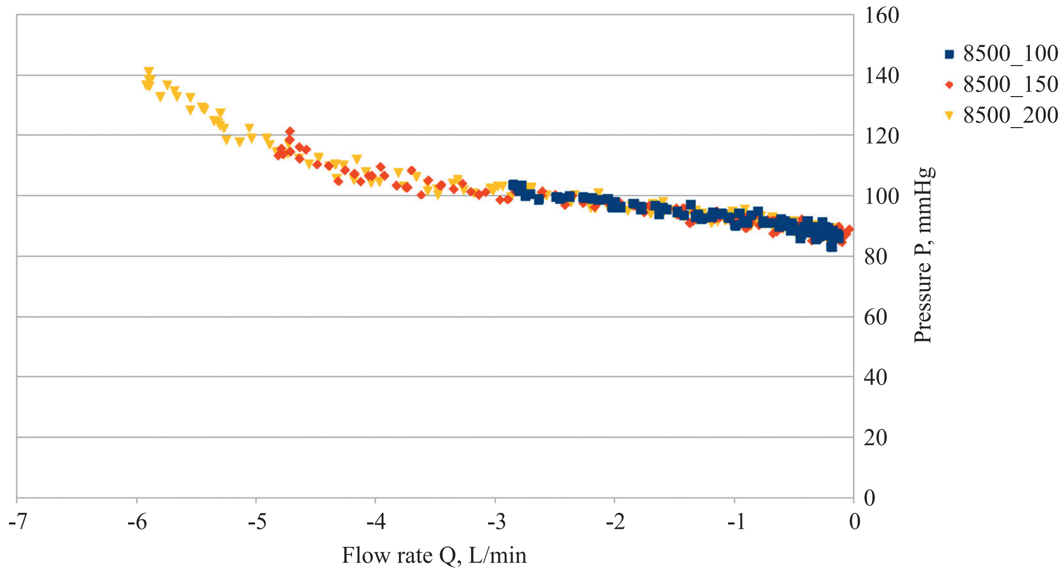

Fig. 6. Head-capacity curves at a rotational speed of 8,500 rpm under different initial pressure conditions

Table

Reverse flow values at different pump rotor speeds

Pump rotor speed (rpm)	Pressure difference P (mm Hg)	Flow rate Q (L/min)
7500	100	–5
7500	90	–4
7500	80	–3
7500	75	–2
7500	70	–1
7500	65	–0.1
8000	110	–5
8000	95	–4
8000	90	–3
8000	85	–2
8000	81	–1
8000	75	–0.1
9000	127	–5
9000	116	–4
9000	112	–3
9000	106	–2
9000	104	–1
9000	97	–0.1
10000	150	–5
10000	140	–4
10000	136	–3
10000	132	–2
10000	124	–1
10000	121	–0.1

The resulting pressure drop and flow rate values obtained at different pump rotor speeds are summarized in Table.

The overall HCCs in the backflow region for all experimental conditions are shown in Fig. 7.

Analysis of the obtained data in comparison with previously established HCCs (Fig. 2) enabled us to determine the total HCC, including the negative flow region, which corresponds to reverse flow conditions.

DISCUSSION

As a result of the conducted experiments, HCCs for an axial-flow pump used in a MCS system were obtained across a range of outlet backpressures and rotor rotational speeds. The shape of the curves in the reverse flow region remains consistent with previously reported HCC behavior. The extent of the negative flow domain was limited by the technical capabilities of the experimental setup.

In the negative flow region, the pressure–flow relationship is preserved, allowing, if necessary, estimation of reverse flow rate (L/min) as a function of rotor speed and system pressure. Thus, the obtained HCC data, including the reverse flow domain, have practical significance and may be used for optimization of operating parameters in MCS systems.

The negative flow rate values obtained in this study represent a continuous extension of the HCC into

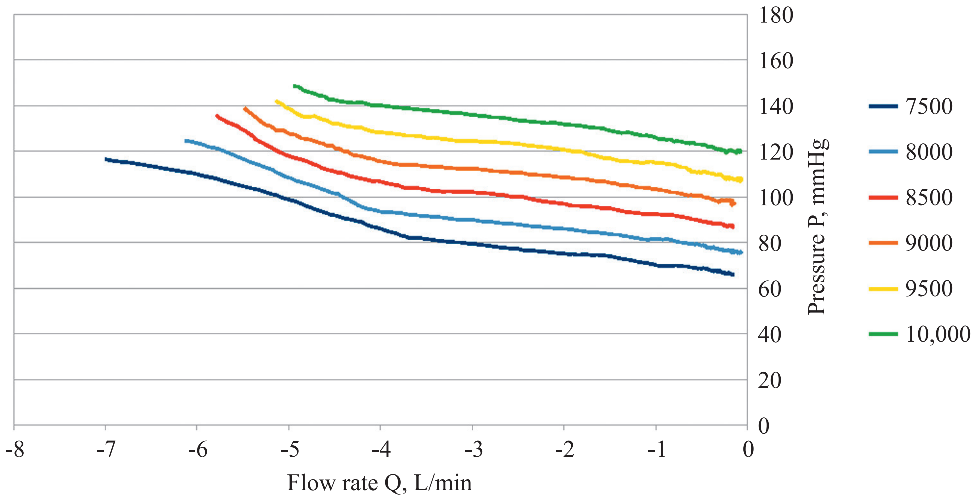

Fig. 7. Head-capacity curves in the negative flow range

the reverse flow region (Fig. 2), while preserving the slope of the characteristic curve. In clinical conditions, particularly during the use of MCS in parallel with a beating heart, the occurrence of retrograde flow from the aorta into the left ventricular cavity during diastole may reduce overall system efficiency. This can result in unpredictable hemodynamic variations and increased risk of complications.

CONCLUSION

The occurrence of backflow in axial-flow pumps used in MCS systems has been identified in the literature as a clinically relevant factor associated with adverse outcomes [4, 7, 10], which is consistent with the findings of the present study.

Despite its clinical significance, reverse flow is not routinely considered when optimizing operating parameters of extracorporeal circulation systems, including widely used clinical devices of various types.

The authors declare no conflict of interest.

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