Hemodynamic Performance of a Novel Right Ventricular Assist Device (PERKAT)

Daniel Kretzschmar,* P. Christian Schulze,* and Markus W. Ferrari†

Acute right ventricular failure (RVF) is an increasing clinical problem and a life-threatening condition. Right ventricular assist devices represent a reasonable treatment option for patients with refractory RVF. We here present a novel percutaneously implantable device for right ventricular support. The PERKAT device is based on a nitinol stent cage, which is covered with valve-carrying foils. A flexible outlet trunk with a pigtail tip is connected to the distal part. The device is driven by an intra-aortic balloon pump (IABP) drive unit, which inflates/deflates a standard IABP-balloon placed within the stent cage. In-vitro evaluation was done in a liquid bath containing water or blood analog. The PERKAT device was tested in different afterload settings using two different IABP-balloons and varying inflation/deflation rates. We detected flow rates ranging from 1.97 to 3.93 L/min depending on the afterload setting, inflation/deflation rate, balloon size, and the medium used. Flow rates between water and blood analog were nearly comparable, and in the higher inflation/deflation rate settings slightly higher with water. Based on this promising *in vitro* data, the innovative percutaneously implantable PERKAT device has a potential to become a therapeutic option for patients with RVF refractory to medical treatment. *ASAIO Journal* 2017; 63:123–127.

Key Words: right ventricular assist device, right heart failure, right ventricular failure, percutanous

The clinical relevance of right ventricular (RV) dysfunction in acute heart failure has gained increasing interest over the past decades. Right ventricular failure (RVF) occurs as a consequence of right ventricular myocyte damage, *e.g.*, after myocardial infarction, because of volume overload, *e.g.*, post-left ventricular assist device-implantation, or because of pressure overload, *e.g.*, after pulmonary embolism.[1] Right ventricular myocardial infarction-associated cardiogenic shock results in a 1 month mortality rate of up to 50%.[2]

Therapeutic management of RV failure includes reversal of the primary cause, inotropic support to enhance cardiac contractility, volume resuscitation to maintain RV preload, and pulmonary vasodilatation to reduce RV afterload.[3] In refractory RV failure, RV assist devices (RVADs) represent a reasonable treatment option. Despite surgical RVADs percutaneously implantable RV support devices illustrate an emerging field. Clinically established strategies such as venoarterial extracorporeal membrane oxygenation, the TandemHeart (CardiacAssist Inc., Pittsburgh, PA),[4] or the Impella RP (Abiomed Inc., Danvers, MA)[5] showed specific limitations such as size, complexity, and absence of a simple percutanous option. RVADs will be needed temporarily and require rapid deployment. We recently published the technical concept and first *in vitro* data of the PERKAT (PERcutaneous KATheterpump) system[6] offering minimal invasive effective RV support. This system is designed for implantation in the inferior vena cava (IVC). As a self-expandable nitinol stent, it only requires 18 French sheath luminal access, and is driven by a standard intra-aortic balloon pump (IABP) console.

The aim of the current study was to evaluate the performance of PERKAT depending in different hemodynamic conditions regarding different viscosities of circulating fluid, various afterload settings, and inflation rates using two different standard IABP-balloon sizes in a standardized *in vitro* model.

Methods

We recently published the concept, technical dossier, and implantation technique of the PERKAT system.[6]

In brief, PERKAT consists of a 220 mm nitinol stent cage encased by flexible membranes containing foil valves. A flexible outlet trunk with a pigtail-shaped tip containing outflow valves is attached to the distal part of the stent cage (**Figure 1**). A standard IABP balloon is placed inside the nitinol stent cage and connected to a standard IABP console as the drive for the PERKAT *via* the helium driveline. Deflation of the IABP generates blood flow into the nitinol stent cage through the foil valves. During balloon inflation, foil valves will be closed because of the pressure. This results in a pulsatile blood flow through the flexible outlet tube towards the pigtail ending. The innovative foil valve concept permits a given direction of blood flow (**Figure 2**). The PERKAT 18 French is designed for percutaneous implantation by Seldinger's technique. The nitinol stent body will be deployed by pulling back the sheath after positioning in the IVC, whereas the outlet trunk bypasses the right atrium and ventricle, and the pigtail tip lies in the pulmonary trunk (**Figure 3**).

From the *Department of Internal Medicine I, Division of Cardiology, Friedrich-Schiller-University Jena, Jena, Germany; and †HSK, Clinic of Internal Medicine I, Helios-Kliniken, Wiesbaden, Germany.

Submitted for consideration April 2016; accepted for publication in revised form October 2016.

Disclosures: This study was supported by the German Ministry of Education and Research (BMBF-PERKAT-13GW0013C). The PERKAT device was developed and designed in cooperation with the Novapump GmbH (Jena, Germany). We thank Ronald Reich, Joerg Pfeifer, Patrick Patzer, and Yu-Jin Heinekamp. The team from Novapump has significantly contributed to the test setup used in the *in vitro* testing of PERKAT and has been working closely with the authors to implement the suggestions for improved handling of the device. Prof. Dr. M.W. Ferrari is a cofounder and a shareholder in Novapump. For the remaining authors, none were declared.

Correspondence: Daniel Kretzschmar, Department of Internal Medicine I, Division of Cardiology, Friedrich-Schiller-University Jena, Erlanger Allee 101, Jena 07747, Germany. Email: daniel.kretzschmar@med.uni-jena.de.

DOI: 10.1097/MAT.0000000000000464

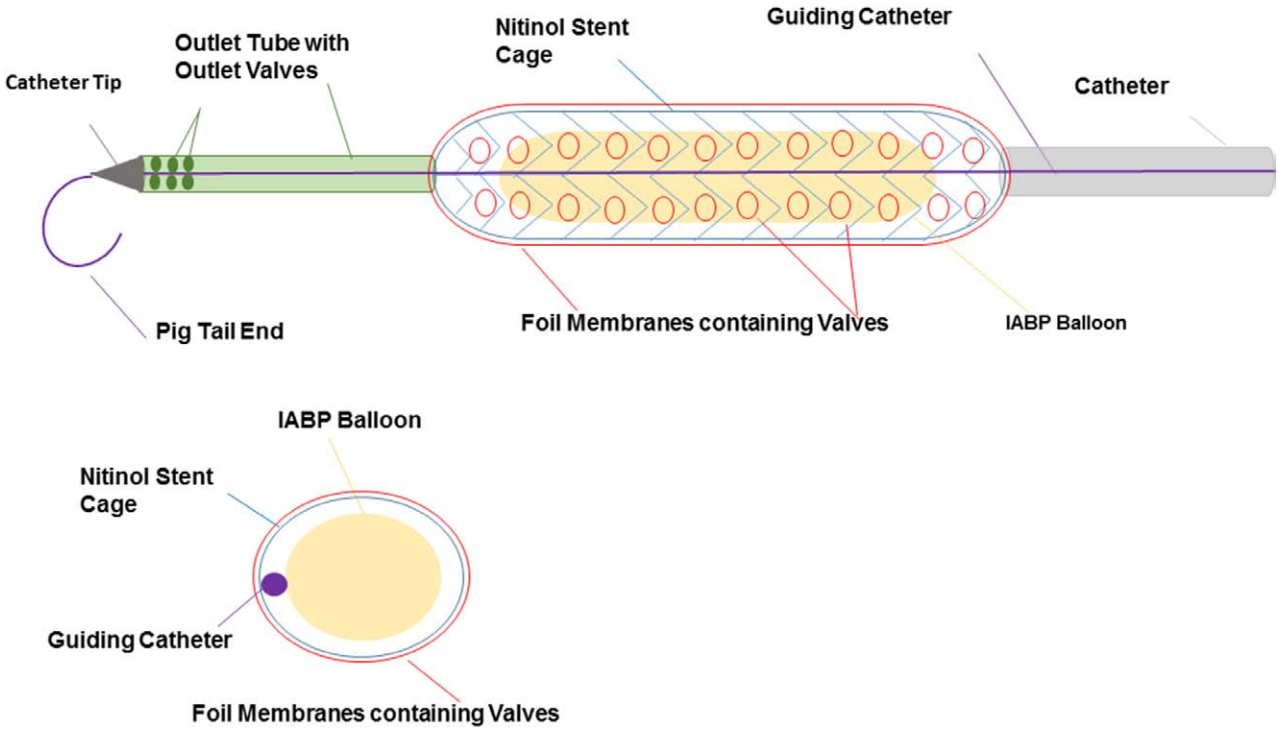

Figure 1. Schematic drawing of PERKAT. IABP, intra-aortic balloon pump. full color online

Composition and working principle of PERKAT is different from that of an IABP. The main difference is the directed flow of blood due to the outlet tube towards the pigtail ending from the IVC straight into the pulmonary arteries without any backflow, whereas the IABP creates an undirected flow with blood also going into the superior vena cava and into the iliac/femoral veins.

In Vitro *Testing*

The lower reservoir with predefined fluid levels simulating the preload contained the PERKAT device (**Figure 4**). The flexible outflow tube of the PERKAT system was attached to the upper reservoir, which was positioned at three specified heights above the lower fluid reservoir to simulate several afterload levels (30 cm = 22 mm Hg; 60 cm = 44 mm Hg; 90 cm = 66 mm Hg).[7] The PERKAT device was then combined with a standard IABP console (*e.g.,* Arrow KAAT II Plus, Teleflex Inc., Morrisville, NC) through the helium drive line. The fluid in the lower reservoir was heated up to 37°C. Then the IABP drive unit was launched and after the initiation process, the experimental setup was ready.

The aim of the setting was to quantify the flow rate of the PERKAT system *via* a magneto-inductive flowmeter (ifm electronics; Type: SM8000), which measured the flow from the upper reservoir back to the lower basin. To obtain the true flow rate during a specific bpm (beats per minute) setting, we measured the flow of PERKAT after the initiation process of the IABP drive unit to avoid start and stop phases.

The flow rate was obtained by measuring the mean flow for 60 sec, three times consecutively. The indicated values represent the mean of those three measurements.

We tested PERKAT in this setting with two different standard IABP balloons (40 ml Sensation Plus and 34 ml Fidelity, Maquet), with varying inflation/deflation rates of 60, 70, 80, 90, 100, 110, and 120 bpm.

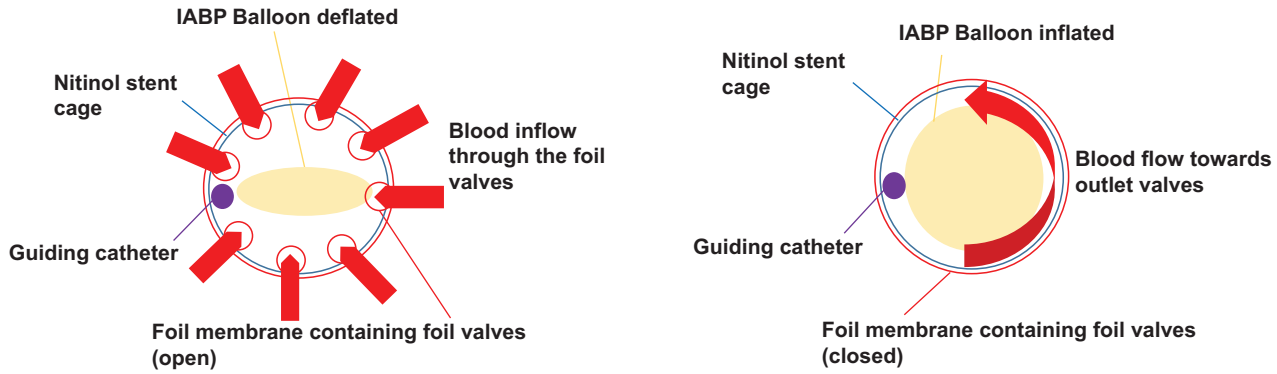

Figure 2. Cross-sectional scheme of PERKAT during IABP-balloon deflation (left) and inflation (right). IABP, intra-aortic balloon pump. full color online

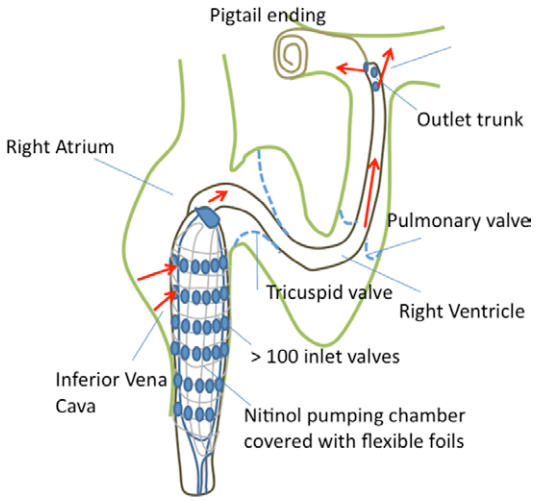

Figure 3. Positioning of PERKAT. full color online

Production of Blood Analog

The viscosity of blood depends on different parameters such as shear rate, hematocrit, and the vessel diameter. For our experimental setup, we aimed to achieve a blood viscosity of 4.07 mPa/sec (at 25°C) and 3.05 mPa/sec (at 37°C), which represents a standard value of blood with a hematocrit of 45%.[8]

The blood analog was produced according to Anastasiou *et al*.[8] and Brookshier and Tarbell.[9] It comprises 74.7% (v/v) distilled water, 25.3% (v/v) glycerol, and 0.032% (w/v) xanthan gum. Xanthan gum is a polysaccharide, which acts as a rheology modifier. The viscosity was determined with the Ubbelohde viscometer according to the operating instructions (Schott). For the production of 8 L blood mimicking fluid according to the mentioned composition, we used 5.98 L distilled water, 2.12 L glycerol, and 2.72 g xantham gum. We measured a viscosity of $\eta = 3.05$ mPa/sec at 37.1°C, according to the literature, this represents the viscosity of whole blood with a hematocrit of 45%. The viscosity of water at 37°C was $\eta = 0.72$ mPa/sec.

Results

In Vitro *Testing with Water*

The detected flow rates are listed in **Table 1**. In all settings, we measured a decrease in flow rates with the increase of the afterload.

In the 22 mm Hg afterload setting, the lowest flow rate of 2.13 L/min was seen with 34 ml balloon at 60 bpm. The highest flow rate of 3.93 L/min was detected at 120 bpm with the 40 ml balloon. Although flow rates were continuously increasing with the 40 ml balloon, we measured the same flow rate with the 34 ml balloon at 110 and 120 bpm.

In the 44 mm Hg setting, the lowest flow rate was measured with the 34 ml balloon at 60 bpm and the highest rate was found at 120 bpm with the 40 ml balloon. Flow rates increased continuously with both balloons.

In the 66 mm Hg setting, the lowest flow of 1.97 L/min was seen at 60 bpm, with the 34 ml balloon, the highest flow was measured with the larger balloon at 120 bpm. Flow rates were constantly increasing with the increase of the inflation/deflation rate in the 34 ml balloon setting. The flow rates with the 40 ml balloon increased from 60 to 110 bpm, while flow rates at 110 and 120 bpm were comparable.

In an additional experiment, we could detect flow rates of up to 3.5 L/min (with each balloon) at an afterload of 88 mm Hg (data not shown).

In all afterload and bpm settings, flow rates with the 40 ml balloon were higher in comparison to the 34 ml balloon.

With increase in the afterload, the flow rates were decreased. This was more pronounced from 22 to 44 mm Hg in comparison with 44 to 66 mm Hg, where flow rates were nearly comparable.

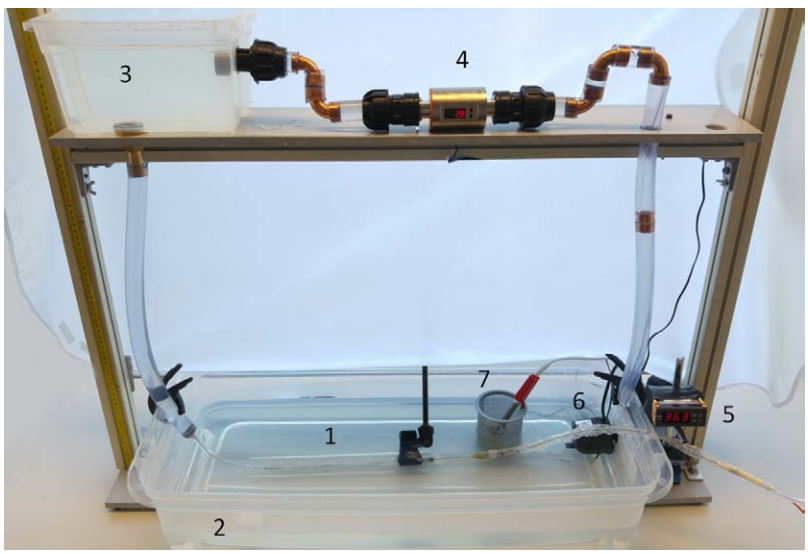

Figure 4. Experimental setup for in-vitro testing: 1, PERKAT; 2, lower reservoir (preload); 3, higher reservoir (afterload; overflow reservoir); 4, magneto-inductive flowmeter (ifm electronics; Type: SM8000); 5, temperature control (Digital-Temperature control WH7016D 24V); 6, circulation pump (EHEIM compact 600); and 7, immersion coil 350 W (ROMMELSBACHER RT 350). full color online

Table 1. Flow Rates for *In Vitro* Testing with Water

Inflation/Deflation Rate (bpm)	22 mm Hg, 34 cc Balloon	22 mm Hg, 40 cc Balloon	44 mm Hg, 34 cc Balloon	44 mm Hg, 40 cc Balloon	66 mm Hg, 34 cc Balloon	66 mm Hg, 40 cc Balloon
60	2.13 ± 0.03	2.33 ± 0.03	2.03 ± 0.03	2.27 ± 0.03	1.97 ± 0.03	2.23 ± 0.03
70	2.47 ± 0.03	2.67 ± 0.03	2.37 ± 0.03	2.63 ± 0.03	2.33 ± 0.03	2.57 ± 0.03
80	2.87 ± 0.03	3.0 ± 0.00	2.70 ± 0.06	3.03 ± 0.03	2.63 ± 0.03	2.93 ± 0.03
90	3.2 ± 0.06	3.43 ± 0.03	3.00 ± 0.06	3.23 ± 0.03	2.97 ± 0.03	3.27 ± 0.03
100	3.47 ± 0.03	3.63 ± 0.03	3.30 ± 0.06	3.53 ± 0.03	3.27 ± 0.03	3,43 ± 0,03
110	3.63 ± 0.03	3.83 ± 0.03	3.43 ± 0.03	3.77 ± 0.03	3.43 ± 0.03	3.63 ± 0.03
120	3.63 ± 0.03	3.93 ± 0.03	3.63 ± 0.03	3.83 ± 0.03	3.53 ± 0.03	3.63 ± 0.03

Values are given as mean ± standard error of the mean.

In Vitro *Testing with Blood Analog*

All measured flow rates are presented in **Table 2**.

With an afterload of 22 mm Hg, the lowest flow rate of 2.13 L/min was seen at 60 bpm with the 34 ml balloon, the highest flow rate was 3.67 L/min at 120 bpm with the 40 ml balloon. With both balloons, we measured continuously rising flow rates, with the increase of the inflation/deflation rate. The performance of the 40 ml balloon was higher than that of the 34 ml balloon in all settings.

In the 44 mm Hg afterload setting, the lowest flow rate of 2.03 L/min again was seen at 60 bpm with the 34 ml balloon, the highest flow rate was 3.50 L/min at 120 bpm with the 40 ml balloon. Flow rates increased from 60 to 120 bpm in both balloon measurement rows with higher flow rates of the 40 ml balloon.

With 66 mm Hg afterload, the lowest flow rate of 1.97 L/min was detected at 60 bpm with the 34 ml balloon, the highest flow rate was 3.47 L/min at 120 bpm with the 40 ml balloon setting. Again, the performance of the 40 ml balloon was higher than that of the 34 ml balloon in all settings with continuously increasing flow rates of the 34 ml balloon with an increase of the inflation/deflation rate, whereas the flow rates of the 40 ml balloon were comparable between 110 and 120 bpm.

In summary, hemodynamic performance was decreasing with increasing afterload settings. The decline in flow rates from 44 to 66 mm Hg was only marginal.

Comparison of Water and Blood Analog

Flow rates of water and blood analog are shown in **Figure 5**.

The flow rates of the 34 ml balloon were comparable in both settings between 60 and 90 bpm, respectively, 100 bpm at an afterload setting of 44 mm Hg, at higher bpm rates, PERKAT performed better in the medium water in comparison to blood analog.

In the 40 ml balloon setting, flow rates were comparable between 60 and 100 bpm. The performance of PERKAT was lower with blood analog at 110 and 120 bpm in comparison to water.

Discussion

We demonstrated in the current study that PERKAT is able to generate a flow rate of up to 3.9 L/min in a standardized *in vitro* model depending on the size of the IABP-balloon, the afterload setting and inflation/deflation frequency. Flow rates obtained with a blood analog were comparable or with increasing inflation/deflation rates slightly lower than the flow rates obtained with water.

We detected the highest flow rates in the 22 mm Hg afterload setting. With increasing afterload, the flow rates were lowered with both fluid media. The decrease was more pronounced with an afterload augmentation between 22 and 44 mm Hg, than from 44 to 66 mm Hg in both media. To some extent, the flow rates measured in the 44 mm Hg were comparable with the rates detected in the 66 mm Hg setting. In both media, we unraveled that the hemodynamic performance of the 40 ml balloon is better than the 34 ml balloon. With the 34 and 40 ml balloon, we could detect a positive correlation between increasing inflation/deflation rates and the resulting flow rate. In the lower range of inflation/deflation frequency (60–100 bpm), the obtained flow rates were comparable between water and the blood analog, irrespective of the used balloon and the afterload setting. As expected in the upper range (100–120 bpm), flow rates measured in blood analog were slightly smaller than in the medium water. We refer this to the higher viscosity of the blood analog in comparison to water. Although we could detect only a slight difference in the hemodynamic performance of PERKAT between water and blood analog, we further recommend the use of a blood analog for device testing because this is stated in the AAMI/ISO standards.

Table 2. Flow Rates for *In Vitro* Testing with Blood Analog

Inflation/Deflation Rate (bpm)	22 mm Hg, 34 cc Balloon	22 mm Hg, 40 cc Balloon	44 mm Hg, 34 cc Balloon	44 mm Hg, 40 cc Balloon	66 mm Hg, 34 cc Balloon	66 mm Hg, 40 cc Balloon
60	2.13 ± 0.03	2.27 ± 0.03	2.03 ± 0.03	2.23 ± 0.03	1.97 ± 0.03	2.27 ± 0.03
70	2.43 ± 0.03	2.63 ± 0.03	2.33 ± 0.03	2.67 ± 0.03	2.33 ± 0.03	2.63 ± 0.03
80	2.87 ± 0.03	2.97 ± 0.03	2.73 ± 0.03	2.93 ± 0.03	2.63 ± 0.03	2.97 ± 0.03
90	3.17 ± 0.03	3.33 ± 0.03	2.97 ± 0.03	3.23 ± 0.03	2.97 ± 0.03	3.23 ± 0.03
100	3.40 ± 0.03	3.6 ± 0.03	3.27 ± 0.03	3.47 ± 0.03	3.13 ± 0.03	3.4 ± 0.00
110	3.53 ± 0.03	3.63 ± 0.00	3.37 ± 0.03	3.47 ± 0.03	3.27 ± 0.03	3.47 ± 0.03
120	3.57 ± 0.03	3.67 ± 0.03	3.47 ± 0.03	3.5 ± 0.00	3.30 ± 0.03	3.47 ± 0.03

Values are given as mean ± standard error of the mean.

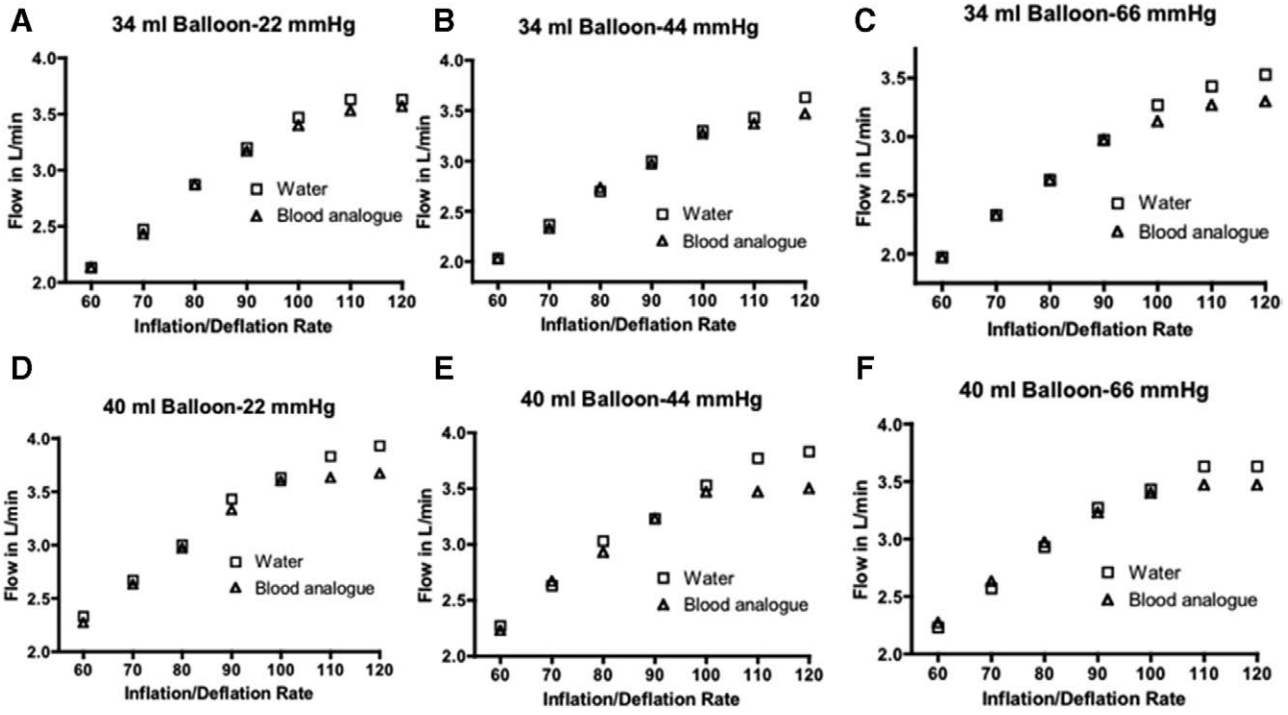

Figure 5. Flow-rate performance of the 34 and 40 ml IABP-balloon using different afterload settings and the media water and blood analog. Flow rate of the 34 ml balloon with 22 mm Hg afterload (**A**); 34 ml balloon with 44 mm Hg afterload (**B**); 34 ml balloon with 66 mm Hg afterload (**C**); 40 ml balloon with 22 mm Hg afterload (**D**); 40 ml balloon with 44 mm Hg afterload (**E**); and 40 ml balloon with 22 mm Hg afterload (**F**). full color online

In comparison to our previously published *in vitro* data,[6] we observed significantly higher flow rates. We refer this to the use of a PERKAT prototype in the former study, which has been replaced by an improved model system. Nevertheless, we have to mention the limited comparability between both studies because of the usage of different balloons (30/40 ml balloons in the previous study *vs.* 34/40 ml balloons in the current report). In addition, a magnetoinductive flowmeter was now used to measure the flow rates, which led to a higher accuracy in comparison to the former method of weighing the medium.

Limitations

Because of the fact that blood viscosity depends on several factors such as hematocrit, a viscosity above 3.05 mPa/sec could influence the flow rate. The data received in this optimized model by using large water reservoirs and continuous backflow resulting in low resistance aspiration and ejection by PERKAT need to be further evaluated in an *in vivo* (animal) model.

In summary, we here provide data of the hemodynamic performance of PERKAT under standardized *in vitro* conditions. The device offers hemodynamic support of up to 3.9 L/min. The flow rates obtained with the blood analog were comparable or with increasing inflation/deflation rates slightly lower than the flow rates with water. These data provide first evidence that hemodynamic support is possible through the use of the PERKAT system under consideration of the afterload setting, the used balloon, and the inflation/deflation rate. Further data will have to be collected and confirmed *in vivo*.

Acknowledgments

The authors thank Ronald Reich, Joerg Pfeifer, Patrick Patzer, and Yu-Jin Heinekamp.

References

1. Kapur NK, Paruchuri V, Jagannathan A, *et al*: Mechanical circulatory support for right ventricular failure. *JACC Heart Fail* 1: 127–134, 2013.
2. Jacobs AK, Leopold JA, Bates E, *et al*: Cardiogenic shock caused by right ventricular infarction: A report from the SHOCK registry. *J Am Coll Cardiol* 41: 1273–1279, 2003.
3. Kapur NK, Bader HB: Percutaneous circulatory assist devices for right ventricular failure. *Intervent Cardiol Clin* 2: 445–456, 2013.
4. Giesler GM, Gomez JS, Letsou G, Vooletich M, Smalling RW: Initial report of percutaneous right ventricular assist for right ventricular shock secondary to right ventricular infarction. *Catheter Cardiovasc Interv* 68: 263–266, 2006.
5. Cheung AW, White CW, Davis MK, Freed DH: Short-term mechanical circulatory support for recovery from acute right ventricular failure: Clinical outcomes. *J Heart Lung Transplant* 33: 794–799, 2014.
6. Kretzschmar D, Lauten A, Ferrari MW: *In vitro* evaluation of a novel pulsatile right heart assist device - the PERKAT system. *Int J Artif Organ* 38: 537–541, 2015.
7. Niclauss L, von Segesser LK: PulseCath iVAC 3L™ hemodynamic performance for simple assisted flow. *Interact Cardiovasc Thorac Surg* 12: 912–913, 2011.
8. Anastasioua AD, Spyrogiannia AS, Koskinasb KC, Giannogloub GD, Parasa SV: Experimental investigation of the flow of a blood analogue fluid in a replica of a bifurcated small artery. *Med Eng Phys* 34: 211–218, 2012.
9. Brookshier KA, Tarbell JM: Evaluation of a transparent blood analog fluid: Aqueous xanthan gum/glycerin. *Biorheology* 30: 107–116, 1993.