JECT. 2002;34:92–100
The Journal of The American Society of Extra-Corporeal Technology

Impella: A Miniaturized Cardiac Support System in an Era of Minimal Invasive Cardiac Surgery

Leen Vercaemst RN, ECP, BCP; Eddy Vandezande, ECP, BCP; Patrick Janssens, ECP, BCP; Taverniers Yvan, RN, ECP, BCP; Demuylder Peter, RN, ECP, BCP; Bart Meyns MD, PhD

Department of Cardiac Surgery, University Hospital Gasthuisberg, Louvain, Belgium

Presented at the 39th International Conference of the American Society of Extra-Corporeal Technology, March 22–25, 2001, Miami, Florida

Abstract: In modern coronary bypass surgery, new objectives have been set based upon a minimal invasive approach: beating heart surgery is the new trend to follow, although this might not be feasible in more complex cases. In these cases, the beating heart could be supported by a mechanical device, preferably a device with minimal invasive features to fit in this new approach. For this purpose, two intravascular blood pumps were developed: the Intracardiac Pump LV for left ventricular support and the Intracardiac pump RV for right ventricular support. (Impella Cardiotechnik, Aachen, Germany) The Impella pumps are rotary blood pumps of the axial flow type and produce 4.2 L/min at physiological pressure differences and a rotational speed of 32,500 rotations/min. These micropumps can widen the indications of beating heart surgery by sustaining hemodynamic stability and protecting the heart from warm ischemia. The current concept is aimed at bridging a procedure. Therefore, the proof of safe duration of usage has not been extended beyond 6 hours. As the pump-flow is based on standard pressure-flow curves for each so-called "performance level" (resulting from in-vitro experiments), an investigation was conducted to compare this relationship in the in-vitro trials with the findings in pump-supported patients undergoing coronary bypass surgery. It could be concluded that the intracardiac pump is efficacious in assisting coronary bypass surgery. **Keywords:** Impella, biventricular support, micro-axial blood pumps, minimal invasive, cardiac surgery. *JECT. 2002;34:92–100.*

In modern coronary bypass surgery, minimal invasive approach became an important consideration in the choice of intervention: off pump coronary bypass (OPCAB), minimal invasive direct coronary artery bypass (MIDCAB), hybrid therapy, computer-assisted interventions, and miniaturized heart-lung systems are all witnessing this new trend. As new technologies have always been anticipated by their need, the development of miniaturized support systems was stimulated to meet this new demand of minimal invasiveness.

For this purpose, two intravascular micro-axial bloodpumps were developed: the Intracardiac Pump LV for left ventricular support and the Intracardiac Pump RV for right ventricular support.

The concept of axial flow pump was invented long ago by Archimedes who lived from 287–212 B.C. With his famous screw, he managed to pump water from deep underground to the surface. More than 2000 years later, in 1975, Richard Wampler rediscovered this idea during a roundtrip in Egypt and integrated it into cardiac surgery with a device called the hemopump. The hemopump was the first micro-axial left ventricular support system and was approved for 7 days support (1). In the mean time, the hemopump is no longer manufactured. With the new microaxial pumps, right ventricular support is also possible.

The Impella blood pumps are approved in Europe for support of procedures (safe duration of 6 hours). One of the goals in the further development of this pump is to extend this maximum duration of support up to 7 days.

As the advantages of axial flow pumps became evident by clinicians, more research centers started developing support systems based on this concept. A few examples are: DeBakey, Sun Waseda, TCI Heartmate II, Jarvik 2000, which are all intended for long-term support

MATERIAL AND METHODS

Device Description

Pumps: The two pumps differ in the design of the impeller, the rotor which rotates within a housing, and therefore in the direction of pumping (Figure 1). Additionally,

Address correspondence to: Leen Vercaemst, RN, ECP, BCP, Dienst Perfusie-OK1, U.Z. Gasthuisberg, Herestraat 49, 3000 Leuven, Belgium. E-mail: leen.vercaemst@uz.kuleuven.ac.be
Received March 25, 2001; accepted November 6, 2001.

Article available at https://ject.edpsciences.org or https://doi.org/10.1051/ject/200234292

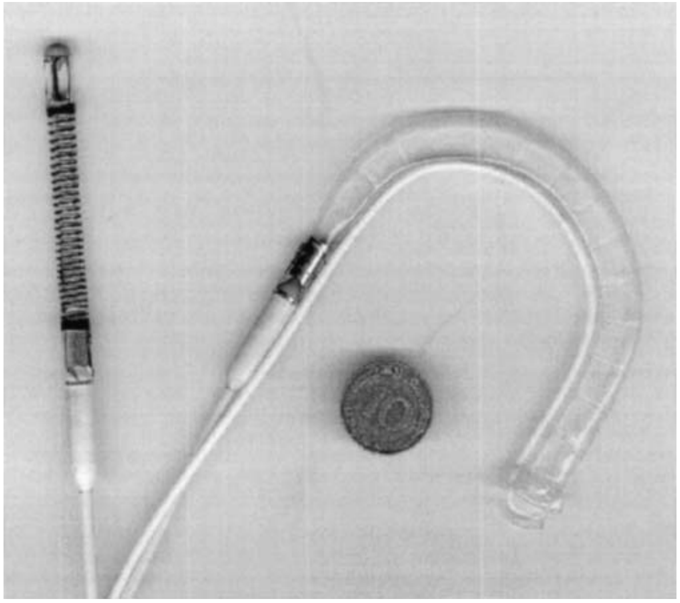

Figure 1. One left and right micro-axial pump.

they differ in the form of the cannulas attached to the pumps, which makes the placement and the accurate pumping position possible in the right and left ventricle.

Both pumps are mounted on an 8 French catheter. This catheter allows integration of the pump into the vascular system, and it contains all wiring for energy support and measuring signals such as rotational speed, differential pressures, and motorcurrent. The pump housing has a diameter of 6.4 mm and includes the impeller and the driving motor.

The differential pressure sensor is located between the canula and the motor. It consists of a diaphragm, which is distorted by pressure applied at its two surfaces (Figure 2). Pressure applied from outside the pump corresponds to the pressure inside the surrounding vessel. Pressure applied to the membrane from within the pump housing corresponds to the pressure at the tip of the canula. Pressure measured by the sensor corresponds to the difference between the two prevailing pressures.

If the left pump has been correctly positioned, differential pressure measured by the sensor can be expressed by the following equation:

$$P\ (\text{diff}) = P\ (\text{aorta}) - P\ (\text{LV})$$

If the right pump has been correctly positioned, differential pressure measured by the sensor can be expressed by the following equation:

$$P\ (\text{diff}) = P\ (\text{pulmonary}) - P\ (\text{RA})$$

This differential pressure sensor allows the monitoring of the pump placement and position as well as the performance. Although the measured signal correlates with the physiological differential pressure, it is influenced by the hemodynamics of the pump (Figure 3).

Cannula: The LV pump has a straight, 6.4 mm wide flexible inflow cannula, which is transvalvular placed through the aortic valve. The differential pressure (difference between aortic and left ventricular pressure) reflects the correct position of the pump in the left ventricle.

The RV pump has a curved, 7.4 mm wide outflow cannula with an inflatable balloon at its tip. Aligned with the cannula, a catheter is located for tip-pressure measurement. The RV pump is placed in a Swann Ganz-like method, indicating that its inflow is located in right atrium and its outflow in the pulmonary artery. The tip of the right cannula has a reverse flow cap that was designed to keep the pump in place during operation. Both pumps can be introduced either peripherally or centrally.

Console: The driving console serves as the interface between the pump and the user. A touch screen displays the differential pressures, pump flows, motor currents, and the right tip pressure. The console allows 2 pumps to be driven at the same time, while simultaneously testing and priming a third pump. This is a redundancy feature in case of a problem with one of the other pumps.

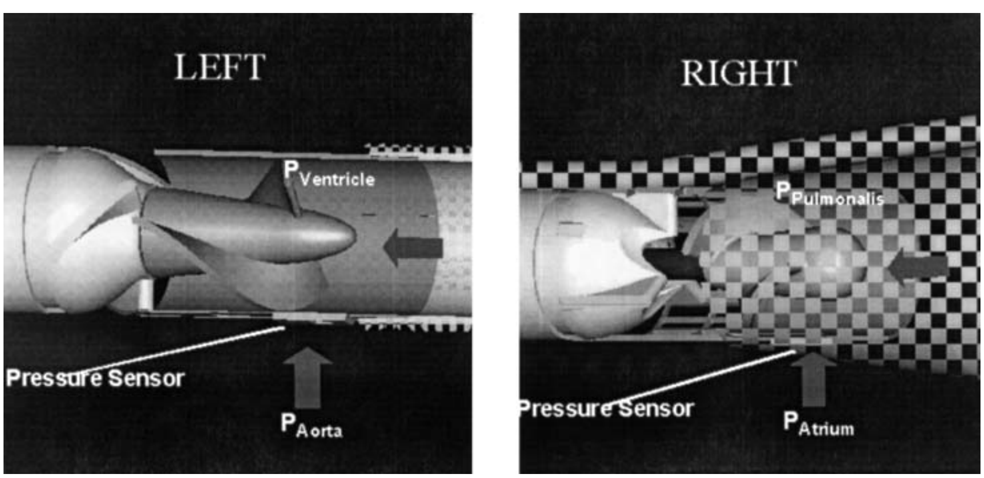

Figure 2. Differential pressure sensor in the left and right pump.

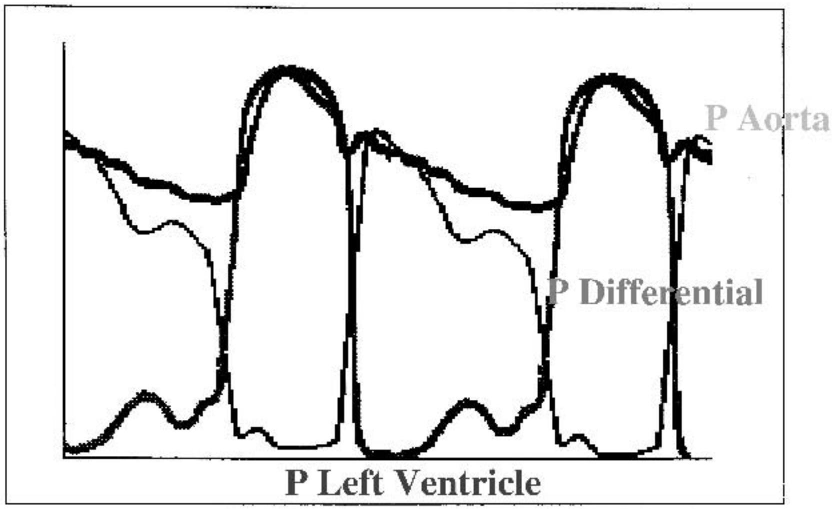

Figure 3. Differential pressure signal in the left pump = P aorta − P left ventricle.

Operation: The intracardiac pump systems are placed transvalvulary (Figure 4), resulting in an interaction of the pumps and the natural heart. The performance of the pumps depends on the difference in pressure before and behind the valve(s) at a certain time.

The left pump aspirates blood from the left ventricle and expels it into the aorta. The right pump is designed with a reversed flow impeller that allows the blood to be pushed forward from the right atrium towards the pulmonary artery.

The pump speed can be adjusted at 10 distinct speed levels (performance 0–9), resulting in a maximum rotational speed of 32,500 rpm and a maximum blood flow of 4.5 L/min. The calculation of the pump flow is based on standard pressure-flow curves for each so-called "performance level" resulting from in-vitro experiments (Figure 4).

The relationship between differential pressure, reproduced flow rates, and performance levels of these micropumps were well investigated during pre-clinical trials. The purpose of this study was to compare this relationship in the in-vitro trials with the findings in pump-supported patients undergoing coronary bypass surgery.

In Vivo Study

The feasibility of this study and the in-vivo analysis of the flow performance were analyzed in a clinical study in 12 patients undergoing coronary artery bypass grafting, (University of Louvain, Belgium and Heart Center at the University of Leipzig, Germany).

The left side pump was inserted either peripherally via femoral artery or centrally by cannulating the ascending aorta. The right side pump was inserted via the superior vena cava.

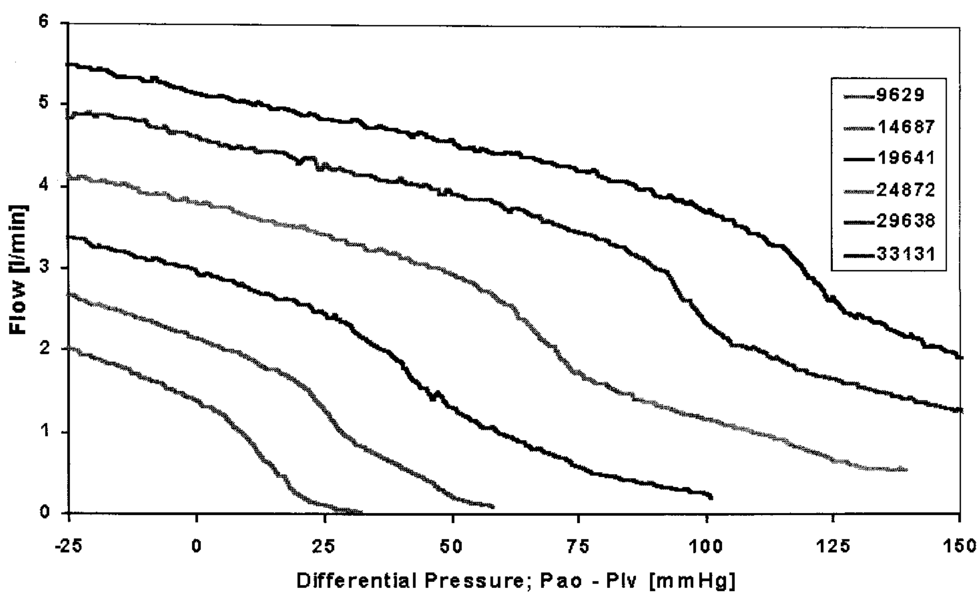

Figure 4. Pressure-flow curves for each performance level in the left pump.

During surgery, all pump data were stored online on a notebook computer. The sampling rate for the pump signals was 25 times/sec. To evaluate the pump data, the pump assistance time was artificially divided into 5-second intervals and for each interval the average of all (5 × 25) pump signals was computed. The interval averages were used as data for further evaluation.

To describe the haemodynamic performance of the pumps during surgery, the mean standard deviation (SD) and differential pressures (minimum and maximum) for each performance level were collected for all patients participating in the clinical trial.

RESULTS

The curve shown in Figure 5 suggests a nonlinear correlation between pump speed and differential pressure measured by the integrated pressure sensor. The pressure signal shows a continuous increase with pump speed level from P1 to P5. The pressure is relatively unchanged between P5 and P7, but shows a marked increase from P8 to P9. The high-pressure values at P9 are partly due to the occurrence of suction.

Similar to the differential pressure curve, the relationship between performance level and pump flow rate suggests being nonlinear (Figure 6). Pump flow increases from P1 to P3 with increasing pump speed. The average flow rate at P4 seems to be lower than at P3. From P4 to P8, there is a continuous increase with pump speed again, followed by a lower average flow at P9 compared to P8. As expected, this corresponds inversely to the pressure signal, which shows a marked increase between P8 and P9.

Characteristic Levels of Differential Pressure and Pump Flow of the Lv Pump

Figure 7 represents the relative occurrence of distinct

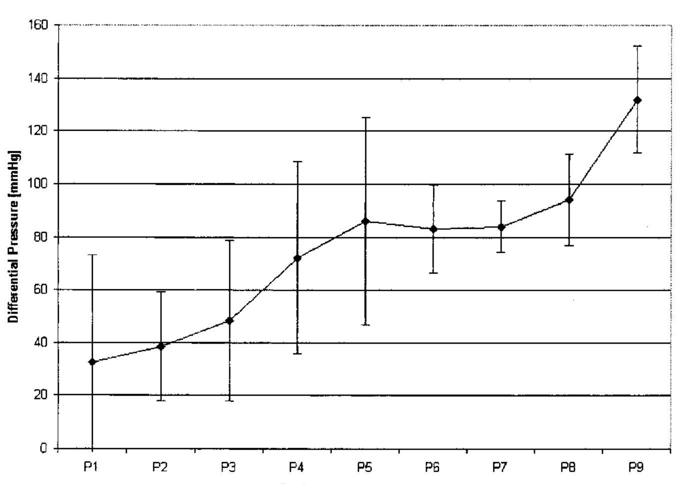

Figure 5. Mean SD maximal and mimimal differential pressure of the 5-second averages of all patients at the various performance levels.

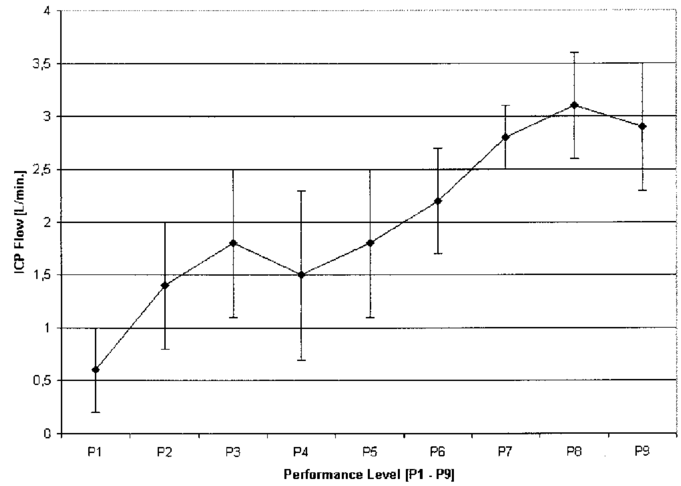

Figure 6. Mean pump flow [L/min] versus performance level of the LV pump.

differential pressure values in percent of total time during which the pump was operated at each performance level. There are one or more peaks for each performance level, which represents the characteristic differential pressure(s) that occur(s).

As seen from the total *n* values (5-second intervals), the LV-pump was mainly operated between performance level P6 and P9. This is because the pump is operated either at P1 to counteract reflux after pump placement or higher levels than P5 to achieve sufficient haemodynamic support.

Performance level P8 shows the highest occurrence of applied operation level. The differential pressure at which the peak of relative occurrence is seen ranges from 10 mmHg at performance level P1 to 120 mmHg at P9. For several pump speeds, there is a second peak with a lower relative occurrence, approximately 20 to 30 mmHg above the first peak.

For each performance level, the relative occurrence of distinct pump flow rate values was calculated in percent of total time during pump run of all patients. There was one or more peaks for each performance level representing the characteristic flow rate that occur. The maximal flow rate was about 4 L/min at a pump speed of 30,000 rpm. As the pump flow rate is a calculated value, similar to the differential pressure signal, the pump flow rate shows one or more sharp peaks for each performance level. The flow rate at which the occurrence has a maximum increase with pump speed ranging from 0.8 to 3.4 L/min at maximum pump speed.

In-vivo Pressure-Flow Curves:

The measured relationship between differential pressure and pump flow values for each performance level was plotted in a scatter plot (Figure 8). As anticipated, the resulting curves resemble the standard curves obtained

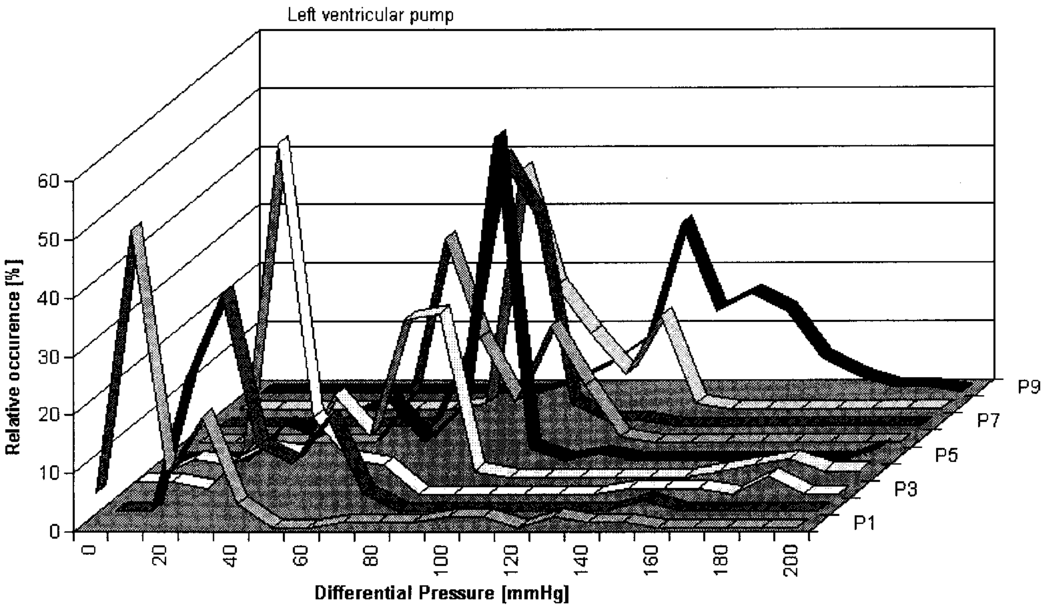

Figure 7. The relative occurrence of distinct differential pressure values, calculated as percentage of total (100%) pump run time in all patients, is presented for each performance level.

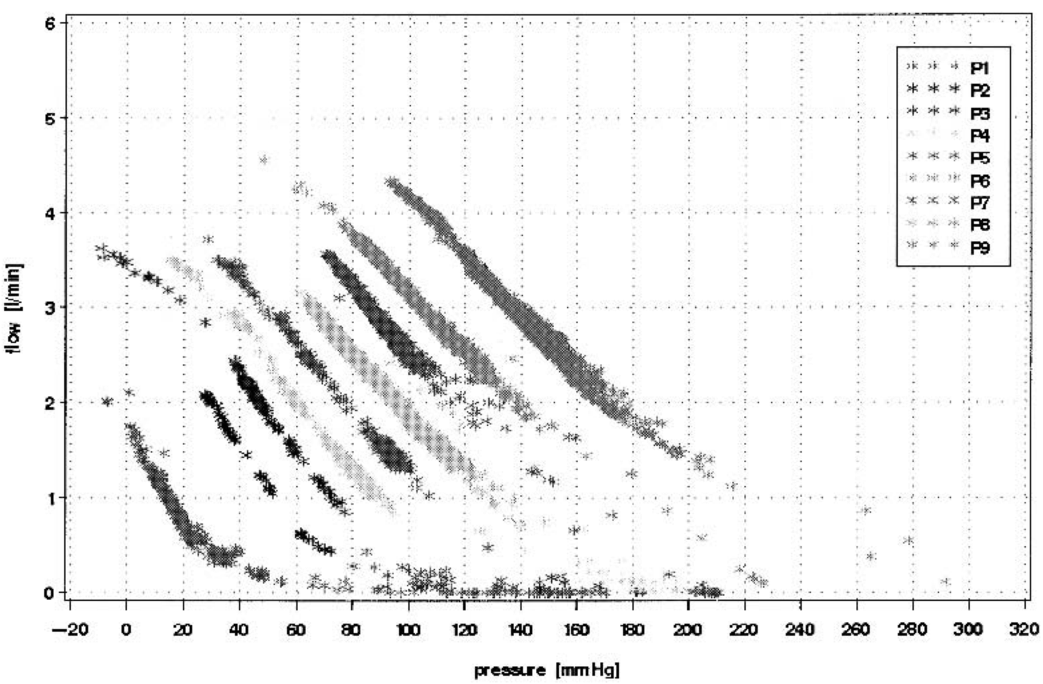

Figure 8. Pump flow rate [L/min] versus differential pressure [mm Hg]. Probability of occurences of measured values of the LV pump.

from in-vivo experiments. However, the scatter plot shows the probability of occurrence of measured values in the clinical trial. The achievable flow rate is dependent to patient blood pressure conditions. However, data points outside the linear range represent unusual operating conditions, such as malposition, suction or kinking of the inflow cannula, or artifacts due to electromagnetic interference caused by electro-surgical instruments. However, the majority of dots (probability of occurrence) for P9 are found to be lower flow than for P8. In the living heart, the differential pressure and flow of the intracardiac pump is influenced by the cardiac function, i.e., the pressure and flow generated by the patient's heart and vascular system.

Plotting the maximum pump flow values versus the minimal differential pressure values within 5-second intervals represents the condition during systole. In this stage, the measured differential pressure solely represents the action of the pump because the aortic valve is open and the pressure difference between the aorta and the left ventricle (Paorta–Pventricle) is at a minimum.

As shown in Figure 9, the flow rates achieved by the pump during the patient's systole are higher than the av-

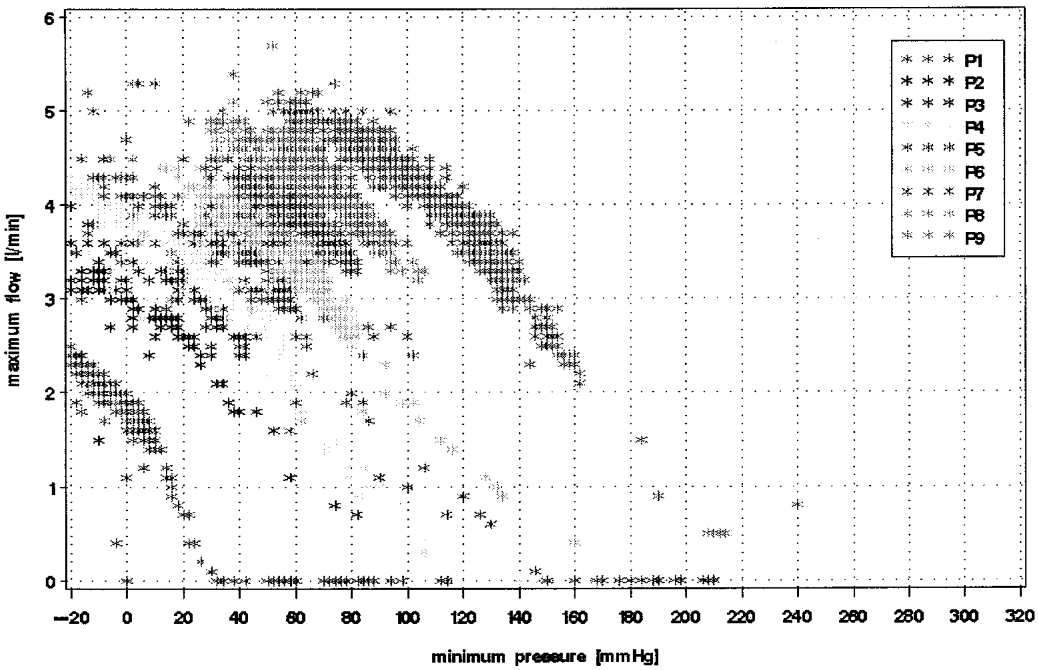

Figure 9. Scatter plot of the occurring relationship between maximal pump flow and minimal differential pressure of all patients during the systole. A single dot represents within a 5-second interval the maximal pump flow and the minimal differential pressure observed.

erage values. Again, there is a characteristic differential pressure-flow relationship for each pump speed. The maximum blood flow at high pump speed is about 5 L/min. However, again the majority of dots (probability of occurrence) for P9 is found to be lower flow than for P8.

Figure 10 represents the condition during diastole when the heart is filling and the aortic valve is closed, and the pressure difference between the aorta and left ventricle (Paorta–Pventricle) is at a maximum and the total blood flow through the pump is at a minimum. The pump solely provides the total cardiac output at this time point. It should be noted that suction occurs mainly during diastole when the mitral valve opens and the lateral leaflet approaches the orifice of the LV pump. Thus, for all performance levels, minimal flow levels down to zero are found.

As shown in Figure 10, the pump flow rates during the patient's diastole are lower compared to the average values, and there is a characteristic differential pressure-flow

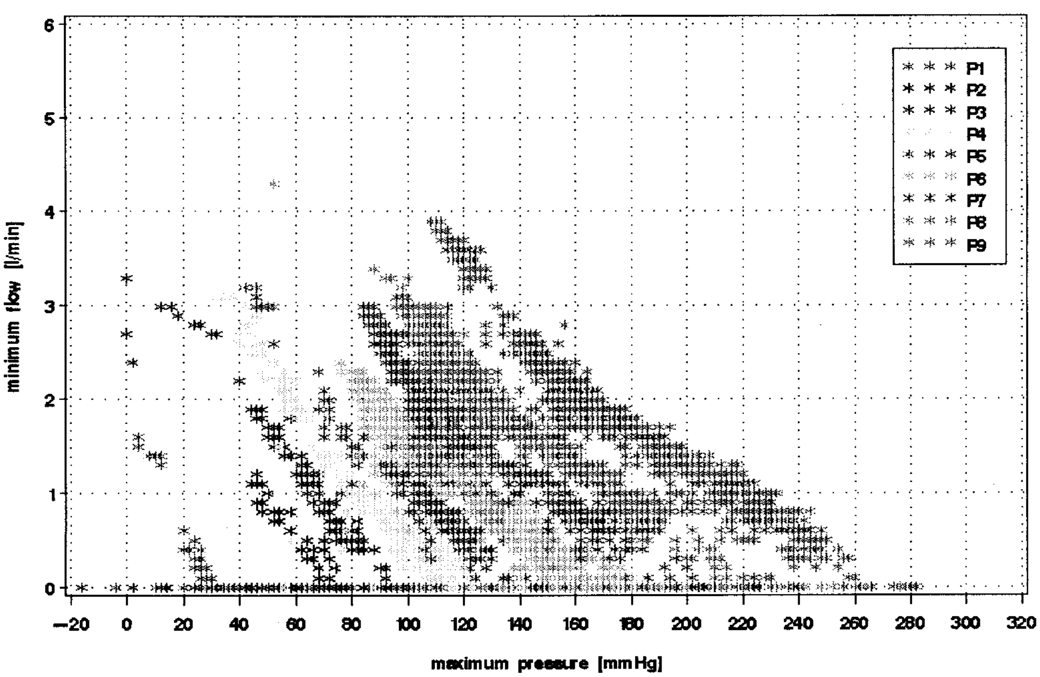

Figure 10. Scatter plot of the occurring relationship between the minima of pump flow and the maxima of differential pressure of all patients during the diastole. A single dot represents within a 5-second interval the minimal pump flow and the maximal differential pressure observed.

rate relationship for each pump speed. The maximum blood flow that can be achieved is approximately 4 L/min.

Right Ventricular Intracardiac Pump (Rv Pump)

Similar to the LV pump, there is an increase in pressure with increasing pump speeds from P1 to P9 and a marked additional increase between P8 and P9, indicating the occurrence of relatively high pressures at maximum pump speed. The high-pressure values may be partially explained by a kinked outflow cannula. The effect of an outflow obstruction may result in a larger pressure increase at maximum pump speed.

It should be noted that suction does not result in a change in differential pressure with the RV pump, contrary to the LV pump. The sensor of the RV pump is less sensitive to suction due to the position of the sensor in relation to the inflow window.

There is a nonlinear correlation between the performance levels and the blood flow rate. The flow increases with pump speed from P1 to P8, followed by a decrease in average flow between P8 and P9. This decrease in flow inversely correlates with the increase in pressure between P8 and P9.

Characteristic Levels of Pressure and Pump Flow in the Rv Pump

For each performance level, the relative occurrences of distinct differential pressure values were calculated in percentage during the pump run of all patients. There are one or more peaks for each performance level, which represents the characteristic pressure level that occurs at each pump speed.

The pump is mainly operated at performance levels between P6 and P9 with a maximum at P8. The occurrence of levels P2 to P5 is very low, indicating that the pump is mainly operated at P1 during standby or higher than level P5 to achieve sufficient haemodynamic support.

These may include malposition or kinking of the outflow cannula. For example, the second peak at 150 mmHg for performance level P9 represents pump operation with a kinked cannula. At performance level P9, there is also a small peak at 290 mmHg. The occurrence of this differential pressure level is explained by pump operation with a totally kinked outflow cannula. These high differential pressure values are only observed at maximum pump speed.

For each performance level, the relative occurrence of distinct pump flow rate values was calculated in percent of total time during pump run of all patients. There are one or more peaks for each performance level representing the characteristic flow rate that occurs. The maximum flow rate was about 4 L/min at a pump speed of 30,000 rpm.

Similar to the differential pressure signal, the pump flow rate shows one or more sharp peaks for each performance level. The flow rate, at which the occurrence has a maximum, increases with pump speed and ranges from 0.4 to 4.0 L/min at the maximal pump speed level.

For some performance levels there are several peaks, indicating there is more than one single characteristic flow rate for these pump speeds. This correlates with the pressure signal, where different characteristic levels were shown for different performance levels as well. This is to be expected because pump flow is a calculated parameter.

For each pump speed, there is a characteristic curve describing the relationship between the flow rate and the performance level; plotting the measured values shows the probability of occurrence. For each pump speed there are also data points outside the linear range that represent unusual operating conditions, such as malposition, kinking of the outflow cannula, or artifacts due to electromagnetic interference caused by electro-surgical instruments.

For performance level P9, there are three distinct operating areas. The first area at 80 mmHg represents an optimal operating condition without any outflow obstruction and flow rates above 4 L/min. The second area at 150 mmHg shows sub optimal operation conditions with partially kinked outflow cannula, where flow rates are reduced to 2.5 to 3.1 L/min. The area at 300 mmHg represents pump operation with totally occluded cannula where only minimal flow can be achieved. This should be avoided by close monitoring of the pressure signal during pump operation.

Due to relatively low-pressure levels that occur in the right heart when compared to the left heart, no comparison of pump characteristics during systole and diastole were made for the RV pump.

DISCUSSION

The efficiency of the Impella pumps in supporting the left and right ventricle was demonstrated during preclinical and clinical trials (2–3). Meyns et al showed in a sheep model that support with these micropumps during consecutive coronary occlusions could lead to a superior haemodynamic state with better myocardial flow and contractility in the reperfusion phase.

The reason for this phenomenon was dually explained. First of all, there was the improved haemodynamic status during the coronary occlusions, since the depression of the myocardial function by snaring the coronaries led to a severe reduction in output and perfusion pressures in the nonsupported sheep. Secondly, the presence of the left pump caused a significant unloading of the left heart during ischaemia. The unloading effect of the left ventricular pump was shown by the reduced stroke work during the pump run. The protective effect of unloading by an axial flow pump during ischaemia has already been shown before with the hemopump (4).

In his animal experiments, Meyns also reported a superior myocardial bloodflow during reperfusion in the supported animals as compared with the control animals. This improved blood flow was present in all regions of the left ventricle as well as the subendocardium and subepicardium.

The value of the right pump is mainly to ensure the blood supply from the right ventricle towards the left heart, especially during kinking and tilting the heart in an effort to reach lateral and posterior located coronary arteries. Meyns demonstrated this in a short additional protocol that was performed in 3 animals: cardiac output was significantly reduced when the heart was kinked (from 3.1 to 1.1 L/min). The initiation of the left side pump had no effect on cardiac output; the initiation of the right side pump restored cardiac output to baseline levels. Grundeman and colleagues also showed in a pig model that mechanical support of the right heart is essential in maintaining a stable hemodynamic situation when lifting the heart (5).

In a multicenter study, conducted in Leipzig, Germany and Louvain, Belgium, this new device for biventricular intracorporeal circulation (ICC) was compared to the conventional heart-lung-machine (HLM) (6). One hundred patients undergoing coronary artery bypass grafting (CABG) surgery were included. The mean arterial pressure was measured and hemolysis was assessed by evaluating plasma free hemoglobin pre-, peri-, and postoperatively.

Tilting of the heart in the intracardiac supported group, for assessment of the posterior wall vessels, could be conducted under stable haemodynamic conditions in all patients. Blood damage was comparable for both study groups, based on free hemoglobin measurements. A limited degree of hemolysis caused by these micropumps was anticipated by its investigators at the time the pumps were designed.

The damage of blood cells is related to sheer stress and the time the cells are exposed to it. The sheer stress produced by the pump at full speed comes to a level where damage of red cells can be anticipated, but remains within acceptable levels due to extremely short exposure time (between 1 and 2 milliseconds). The theoretical arithmetic possibility of cell damage by passing the impeller lies at 0.0167%. The measured results of the in-vivo and in-vitro experiments demonstrate acceptable hemolysis rates for acute pump support.

The analysis of the haemodynamic data maintained during clinical use confirms preclinical findings. The intracardiac pump system was able to maintain haemodynamic stability throughout the procedure in all cases of beating heart surgery. The integrated pressure sensor allows monitoring of pump placement, pump position during surgery, and recognition of critical operation conditions.

The lower flow rate value (Figure 6) at P4 relative to P3 can be explained by the fact that the pump console is operating in two driving modes because it is difficult to operate electrical micro motors at low speed. As the flow is a calculated value, the change of the driving mode between P3 and P4 results in an artificial step up in the pump flow values.

The decrease in flow rate at P9, compared to P8, is due to the increased likelihood of suction. The fact that P9 was used less often than P8 (Figure 7) may be explained by the fact that suction is more likely to occur at maximum pump speed. In this situation, the pump speed is usually reduced until suction disappears.

For most pump speed levels, there exists a sharp peak indicating that there is a characteristic differential pressure for each pump speed. However, the differential pressure is also influenced by the haemodynamic status of the patient, i.e., left ventricular pressure (pre-load), aortic pressure, myocardial function, as well as operating conditions of the pump itself, i.e., pump position, kinked cannula, suction, and so on.

The second peak seen may represent a second characteristic pressure level for the corresponding pump speeds, respectively, or may indicate a sub-optimal operating condition, which is likely to occur. This includes malpositioning and kinking of the inflow cannula.

Similar to the LV pump, there is a sharp maximum in pressure level for most pump speeds in the RV pump, indicating there exists a characteristic differential pressure for each performance level, ranging from 10 mmHg at P1 to 150 mmHg at P9. The differential pressure of the pump is influenced by the haemodynamic status of the patient, i.e., left ventricular pressure (pre-load), aortic pressure (after-load), myocardial function, as well as operating conditions of the pump itself, i.e., pump position, kinked cannula, suction, and so on.

For several pump speeds, there is a second maximum with a lower relative occurrence approximately 20 to 30 mmHg above the first maximum. This may represent a second characteristic pressure level for the corresponding pump speeds to occur or indicate a sub optimal operating condition, which is likely to occur.

The in-vivo results indicate that at the maximal performance level (P9), the pumps do not necessarily provide maximal flow due to an increased likelihood of suction. Therefore, the pump should be monitored closely to avoid pump operation at sub-optimal or critical status.

Adding up all findings, it can be assumed that these blood pumps could contribute towards widening the indications of beating heart surgery by sustaining haemodynamic stability, protecting the heart from warm ischaemia (left support), and allowing a certain degree of heart manipulation (right support).

Since off-pump surgery is not yet the worldwide standard procedure in coronary bypass surgery, more investigations could be made towards comparing microaxial pump-supported CABG surgery with CABG surgery with heart-lung machine. Another patient group, which could benefit from this type of support, is those undergoing high risk PTCA (left support).

This technology can be upgraded towards further miniaturization (transcutaneous placement) and durability (medium to long-term support in left and right ventricular failure). Miniaturization does not only implement a minimal invasive mechanical support system, but also a rapid and easy introducable mechanical support system, ideally to be placed in acute situations.

A medium-term support system could be placed during high-risk procedures and subsequently be left in place awaiting fully ventricular recovery or even transplantation.

It can be concluded from current findings that the intracardiac pump is efficacious in assisting coronary bypass surgery and that this concept of support could, if further upgraded, lead to a wider range of applications.

REFERENCES

1. Lewis T, Graham TR. Great Britain mechanical circulatory support. p. 136–45.
2. Meyns B, Sergeant P, Nishida T, Perek B, Zietkiewics M, Flameng W. Micropumps to support the heart during CABG. Eur J Cardiothorac Surg. 2000;17:169–74.
3. Meyns B, Sergeant P, Siess T, Zietkiewics M, Perek B, Nishida Y, Flameng W. Coronary artery bypass graft with biventricular microaxial pumps. Perfusion. 1999;14:287–90.
4. Merhige ME, Smalling RW, Cassidy D, et al. Effect of the hemopump left ventricular assist device on regional myocardial perfusion and function. Circulation. 1989;80:158–66.
5. Grundeman PF, Borst C, Verlan CW, Meijburg H, Moües CM, Jansen EW. Exposure of circumflex branches in the tilted, beating porcine: echografic evidence of right ventricular deformation and the effect of right or left bypass. J Thorac Cardiovasc Surg. 1999;118:316–23.
6. Autschbach R, Engel M, Gummert JF, Rauch T, Mohr FW, Böning A, Hamann S, Konertz W, Matschke K, Schöndube F, Wiebe K, Meyns B. A new intracardiac microaxial pump-first results of a multicentre study-presented at 8th congress of the International Society for Rotary Blood Pumps. 2000.