Artificial Organs
18(7):494–499, Blackwell Scientific Publications, Inc., Boston

In Vitro Evaluation of the Influence of Pulsatile Intraventricular Pumping on Ventricular Pressure Patterns

Gerhard Rakhorst, Gijsbertus J. Verkerke, *Ab G. Hensens, *Ebo D. de Muinck, *Paul K. Blanksma, †Massimo Pillon, and †Marcel Jufer

*Center for Biomedical Technology, University of Groningen, Groningen; *Department of Thoracic Surgery, University Hospital, Nijmegen, The Netherlands; and †Swiss Federal Institute of Technology, Lausanne, Switzerland*

Abstract: The Pulsatile catheter (PUCA) pump consists of a single port membrane pump connected to an indwelling valved catheter. This so-called transarterial blood pump was originally designed to be introduced through a superficial artery into the left ventricular cavity to pump blood from the left ventricle into the ascending aorta. By introducing the catheter directly into the thoracic aorta or the pulmonary artery, the possibility is created of applying large-diameter catheter PUCA pumps as left, right, or biventricular assist devices (LVAD, RVAD, or BIVAD) without damaging any of the structures of the heart. The pump performance of an 8 mm PUCA pump prototype (internal diameter catheter, 8 mm; catheter length, 40 cm; stroke volume, 80 ml) was studied in a mock circulation to investigate the influence of pulsatile intraventricular pumping on ventricular pressure patterns. The pumping mode of the PUCA pump was changed from approximately 1:1 ([n + 1]:n) to 1:2 ([½n + 1]:n) and 1:3 ([⅓n + 1]:n) in relation to the frequency of a ventricle-simulating membrane pump. Apart from the pumping mode, timing of the PUCA pump driving system (ejection phase) seems to be crucial in obtaining optimal unloading of the ventricle. **Key Words:** Left ventricular assist device—Counter pulsation—Catheter pump—Temporary mechanical support—Mock circulation.

The pulsatile catheter (PUCA) pump consists of a membrane pump connected to a valved catheter (1). When the catheter tip of the PUCA pump is placed in the left ventricular (LV) cavity, its built in inflow and outflow valves allow blood to be pumped from the left ventricle (via the paracorporeally placed membrane pump) into the ascending aorta (Fig. 1). The pump was originally designed to be introduced into the circulatory system through a peripheral artery (axillary or femoral artery). In this application, the generated flow is mainly limited by the internal diameter of the catheter.

On the basis of studies with closed chest by-pass devices (2,3) and intraventricular blood pumps (4) that used catheters or cannulas with external diameters of 18 to 21 French gauge, the first PUCA pump model had the following configuration: internal diameter (ID) of catheter, 6 mm; catheter length, 120 cm; and pump volume, 100 ml. Postmortem studies of patients who died from cardiovascular diseases demonstrated that atherosclerotic processes narrowed the internal diameter of most peripheral arteries. As a consequence, even very small diameter catheter blood pumps (e.g., 14 French gauge Hemopump; Johnson & Johnson) may require special surgical techniques to enable introduction of the catheter into the systemic circulation.

We therefore decided to evaluate the PUCA pump concept in open chest animal experiments. Open chest experiments have the advantage that large diameter catheters can be used. Also, under these conditions the PUCA pump can be applied as a left, right, or biventricular assist device (LVAD, RVAD, BIVAD) without damaging any of the structures of the heart tissue (i.e., no atrial or ventricular anastomoses).

This article describes in vitro studies with an 8

Received January 1994.
Address correspondence and reprint requests to Dr. G. Rakhorst at Centre for Biomedical Technology, University of Groningen, Oostersingel 59, 9713 EZ Groningen, The Netherlands.

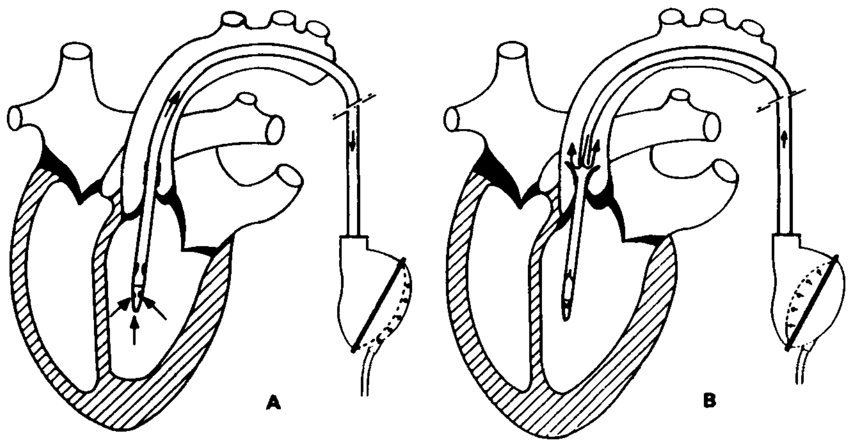

FIG. 1. The PUCA pump concept is shown. The outflow valves are positioned in the near distance of the coronary arteries.

mm (ID) PUCA pump performed to evaluate ventricular and aortic pressure patterns during pulsatile intraventricular pumping.

MATERIAL AND METHODS

PUCA pump configuration

The PUCA pump was made of polyvinyl chloride (PVC) tubing (ID, 8 mm; length, 40 cm) connected to a single port polyurethane membrane pump with a stroke volume of 80 ml (Helmholtz Institute Aachen [HIA], Germany). The inflow opening of the tube was protected against obstructions by a metal cage. A disc valve was placed in the tip of the tube adjacent to the inflow opening. Two outflow openings, with a total surface area greater than the surface area of the inflow opening, were located 15 cm from the inflow valve. These openings were covered by a thin polyurethane sheet to create a one-way flow. The membrane pump was actuated by a UTAH Pneumatic Heart Driver or by an experimental electrohydraulic actuator specially designed for this occasion by the Swiss Federal Institute of Technology (Ecole Polytechnique Fédérale de Lausanne, EPFL). The following pumping frequencies were used: 46, 23, and 16 bpm.

Mock circulation

The mock circulation (Fig. 2) consists of a reservoir (l × w × h = 400 × 250 × 250 mm) filled with water and two Plexiglas columns (l × w × h = 250 × 250 × 400 mm) placed on top of it. One of these columns simulates the left atrium. This column is continuously filled with water from a water pump placed in the reservoir. If the water column rises above an adjustable level, excess water flows away via a tube to the reservoir. The left atrial pressure (water column) was kept at a constant level of 10 cm H_2O (preload). The second column simulates the arterial system. Vascular compliance was simulated by compressed air located above the water level of the arterial column.

The pressure of the arterial system was kept at 120 mm Hg with the help of a pressure safety valve. The water that leaves this valve is guided to the reservoir. The atrial and the arterial systems are

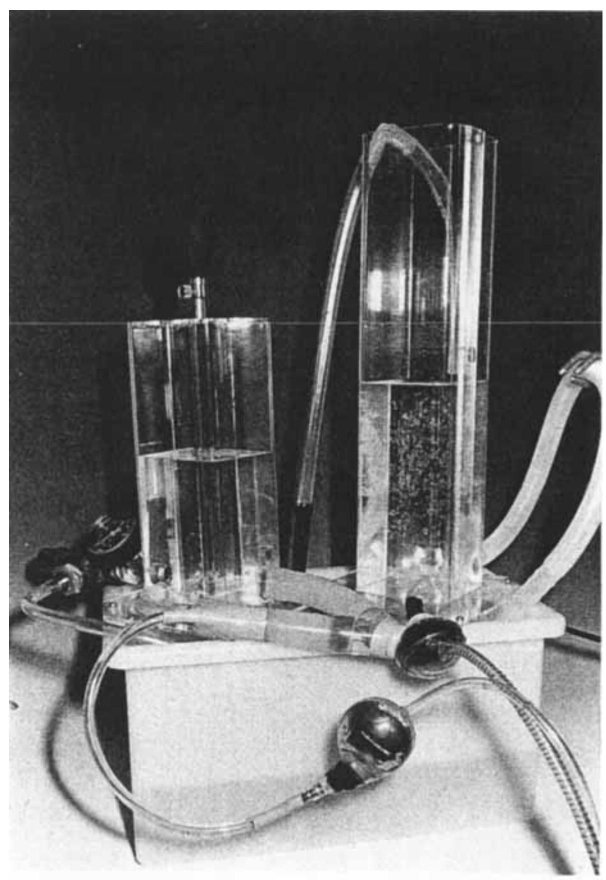

FIG. 2. The PUCA pump mock circulation is shown. The catheter connected to the single port VAD is placed through the outflow valves of the dual-port VAD.

connected by a dual-port membrane pump (HIA 90 ml LVAD) that simulates the natural left ventricle. This membrane pump was activated by the Lausanne driver, and was kept at a pumping frequency of 45 bpm, or if the Lausanne driver was used for the PUCA pump, at the same frequency by the UTAH Pneumatic Heart Driver. Thus, with the PUCA pump frequencies of 46, 23, and 16 bpm, pumping modes of 1:1 ($[n + 1]:n$), 1:2 ($[\frac{1}{2}n + 1]:n$), and 1:3 ($[\frac{1}{3}n + 1]:n$) were established (n = bpm). The tip of the valved catheter of the single port PUCA pump was placed in the outflow tract through the outflow valves of the dual-port membrane pump. The flow generated by the artificial ventricle (aortic flow) and by the PUCA pump was measured separately with electromagnetic flow probes. Pressure transducers were placed inside the artificial ventricle and the ventricular outflow tract (aorta). When atrial or arterial pressures were set, compliance was adapted so that pressure recordings showed physiological patterns.

EPFL experimental driving system

The experimental activator developed by the EPFL is driven by a steppen motor. Rotation to linear movement conversion is done by a roller screw, which has around 90% efficiency and an axial precision lower than 1 μm. The nut of this screw is locked in a linearly guided piston. The leakage of fluid is prevented by special high-performance seals. The position of the piston is examined by optical sensors.

The control system was made by a processor card including a Motorola 68000, which communicates with specific hardware devices. A serial port linked to a terminal with a keyboard permits input of profile parameters and management of the device. System inputs are signals coming from the optical sensors and from a numerical signal to synchronize the cycle. The outputs are composed by an analog signal that corresponds with the flow (maximum flow, 3 L/min) or volume of the pump, as well as by some signals to supply steppen motor phases. The activator can generate a pulsatile functioning mode, with controllable trapezoidal flow profiles.

RESULTS

Ventricular pressure patterns

At equal flow, the pressure patterns of pneumatically and electrohydraulically actuated ventricle-simulating membrane pumps were quite different (Fig. 3). The pneumatic system generated cycles with steep systolic pressure peaks (170 mm Hg) followed by almost horizontal pressure plateaus (120 mm Hg). The pressure patterns of the electrohydraulically activated ventricle were trapezoidal showed systolic pressure peaks of 100 mm Hg.

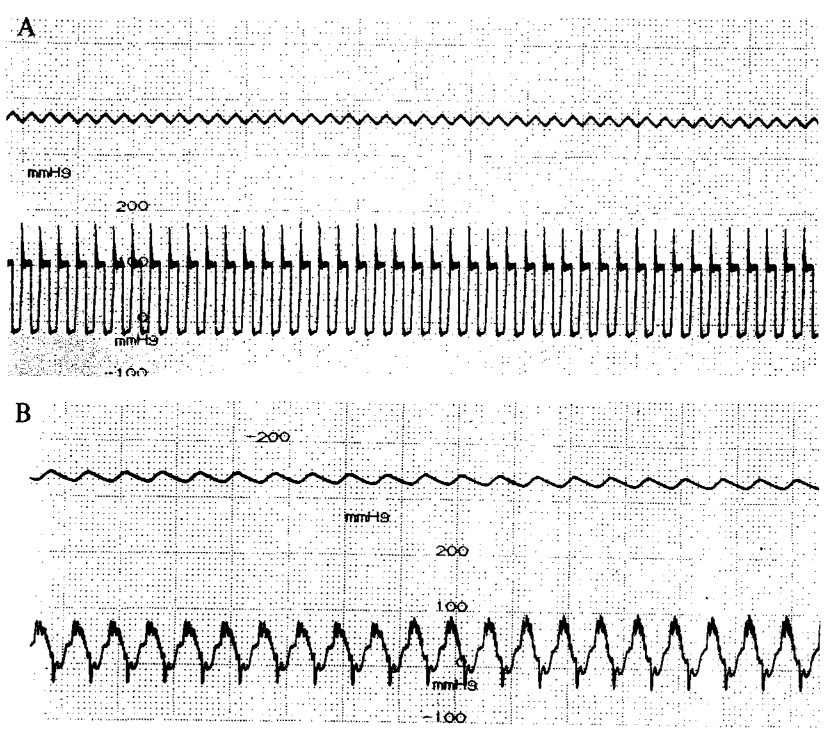

FIG. 3. (A) Ventricular pressure pattern of a pneumatically driven mock circulation (paper speed, 2.5 mm/s) is shown. **(B)** Ventricular pressure pattern of an electrohydraulically driven mock circulation (paper speed, 5 mm/s) is shown.

Pumping mode 46:45

During the pumping mode of (n + 1):n, the pneumatically driven dual-port ventricle demonstrated decreased systolic pressure peaks (100–140 mm Hg) compared with nonassisted pumping. Also, during a 1-min pumping period, the different pump frequencies resulted in periods in which the artificial ventricle was unloaded (low pressure peaks and low negative pressure) and periods in which the artificial ventricle became extra loaded, characterized by high systolic pressure peaks and high negative presystolic pressures (Fig. 4).

During optimal unloading, the arterial (outflow tract) pressure was increased, compared with periods in which extra loading (high ventricular systolic pressures) was registered. This phenomenon was not present during electrohydraulic activation. When the simulated ventricle was electrohydraulically activated, periods of unloading and extra loading became more evident. Optimal unloading resulted in ventricular pressures that were one third of those of the extra loading periods (80–240 mm Hg).

Pumping mode 23:45

With pneumatic activation and a pumping mode of (½n + 1):n, ventricular pressures showed alternating pressure peaks while the ventricular pressure plateau remained at the same level. The unloading and extra loading effects of assisted circulation were less obviously present in the ventricular pressure curves in comparison to the aortic pressure curves of the 1:1 mode. During electrohydraulic activation of the simulated ventricle, periods of optimal unloading could be distinguished from periods of extra loading. However, every ventricular cycle that showed high pressure peaks was followed by a cycle with decreased pressures (Fig. 5).

Pumping mode 16:45

During ventricular assist, the systolic pressure plateau barely changed in the pneumatically driven ventricle. The positive pressure peaks were followed by 2 small peaks, and negative pressure peaks were seen at every third beat. The ventricular pressure pattern of the electrohydraulically driven ventricle shows high systolic pressures followed by 2 beats with low (50% less) systolic pressures. Optimal timing also influences the effects of unloading and extra loading (Fig. 6).

DISCUSSION

This study was meant to investigate the influence of intraventricular pumping on ventricular pressure

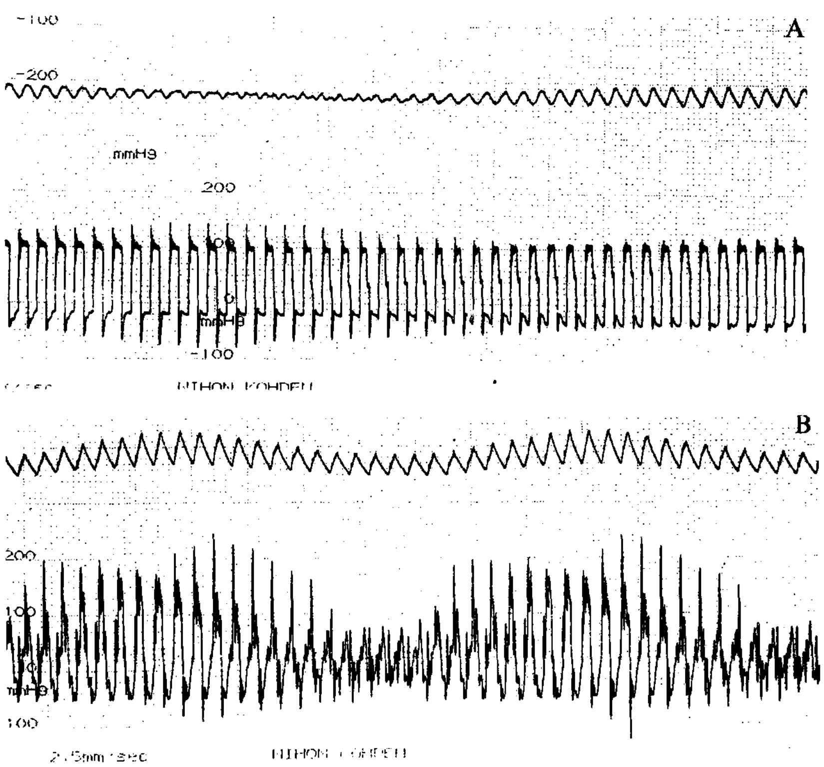

FIG. 4. (A) Ventricular pressure pattern during PUCA assist is shown. Pump mode is 46:45 with a pneumatic driver. **(B)** Ventricular pressure pattern during PUCA assist is shown. Pump mode is 46:45 with an electrohydraulic driver.

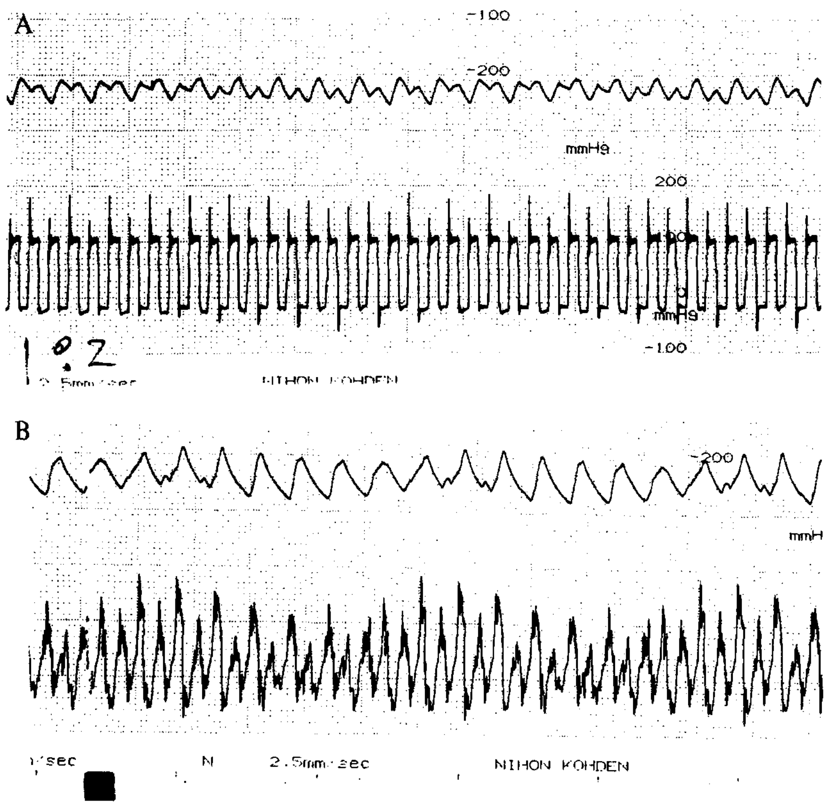

FIG. 5. (A) Ventricular pressure pattern during PUCA assist is shown. Pump mode is 23:45 with a pneumatic driver. **(B)** Ventricular pressure pattern during PUCA assist is shown. Pump mode is 23:45 with an electrohydraulic driver.

patterns. The effect of the PUCA pump on ventricular pressures was evident in either pneumatically or electrohydraulically driven mock circulations. The wave form as a result of optimal timing was clearly present in the pumping mode of 1:1, but was seen at other pumping modes as well. Neither the wave form nor the pumping mode–related ventricular pressure patterns were present in studies with nonpulsatile intraventricular blood pumps (2,5).

Until we exchanged the driving systems of the mock system and the PUCA pump, we were not aware that ventricular performance in mock circulations are so strongly influenced by the driving medium. Mock circulation studies with the Hemopump (5) showed that the influence of intraventricular pumping on ventricular pressure patterns was hard to measure in a pneumatically driven mock circulation. Animal experiments will clarify whether electrohydraulically driven mock circulations are more realistic than pneumatically driven mock circulations.

In our opinion, pneumatically driven VADs in mock circulations will enforce rather constant LV pressures caused by the driving mechanism, and respond little to volume changes in the LV pump, which is not realistic. The negative pressure changes are more pronounced, probably caused by the limited negative pressure capacity of the pressure line.

The experimental Lausanne driver was built to actuate the PUCA pump. In this study, we used the driver to activate a single-port PUCA pump with an 8-mm ID catheter and a ventricle-simulating dual-port HIA LVAD. Due to the limited output of the experimental driver (3 L/min), the flow of the LV in the mock circulation was kept between 2 and 2.5 L/min to enable exchange of both driving systems. Animal experiments will clarify whether flow and pump frequency will influence unloading and extra loading effects of the PUCA pump.

At optimal timing (during periods of maximum unloading), pumping modes of $\frac{1}{2}n + x$ and $\frac{1}{3}n + x$ (x = delay in m/s) may contribute to the counterpulsation effect of the PUCA pump. In this case, every one or two unloaded heart actions will be followed by an extra loaded action. Animal studies are needed to study this effect in more detail.

CONCLUSION

In vitro measurements demonstrated that the PUCA pump possesses specific pump characteristics: it combines intraventricular pumping (ventric-

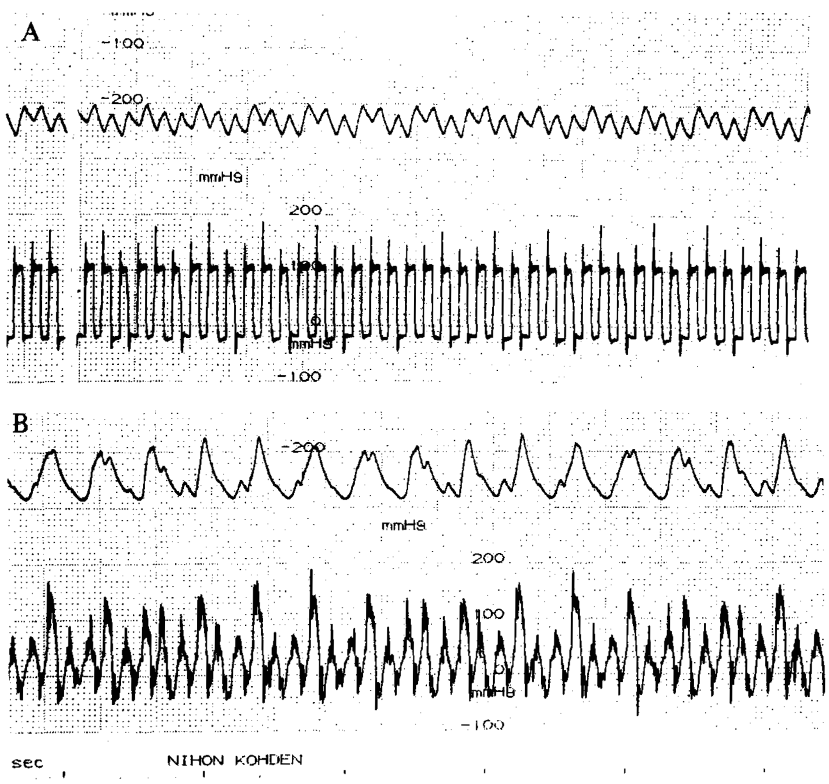

FIG. 6. (A) Ventricular pressure pattern during PUCA assist is shown. Pump mode is 16:45, with a pneumatic driver. **(B)** Ventricular pressure pattern during PUCA assist is shown. Pump mode is 16:45 with an electrohydraulic driver.

ular unloading) with counter pulsation. The effect of intraventricular pumping depended strongly on pump timing. Generated signals were remarkably influenced by the driving systems used (pneumatic or electrohydraulic). This finding may have consequences for the development of future mock circulations.

Acknowledgments: The authors would like to express their gratitude to Prof. Dr. E. Hennig (Berlin Heart Centre) and Prof. Dr. H. Reul (Helmholtz Institute Aachen) for willingness to provide us the essential components of the PUCA pump prototypes. We also thank Hilde Seip and Adam Deerman, students of the Groningen State Polytechniques, for their technical assistance.

REFERENCES

1. Verkerke B, De Muinck ED, Rakhorst G and Blanksma PK. The PUCA pump. A left ventricular assist device. *Artif Organs* 1993;17:365–8.
2. Zwart HHJ, Kralios A, Collan R, Kolff WJ. Transarterial Closed-chest left ventricular (TaCLV) bypass. *ASAIO Trans* 1969;15:386–90.
3. De Paulis R, Engelhardt H, Chiariello L, Reul H, Morea M. In vitro evaluation of left ventricular assistance by cannulation of both femoral arteries. *Int J Artif Organs* 1990;13: 237–46.
4. Wampler RK, Moise JC, Frazier OH, Olsen DB. In vivo evaluation of a peripheral vascular access axial flow blood pump. *ASAIO Trans* 1988;34:450–4.
5. Kunst EE. Quantification of left ventricular function during cardiac assist. *PhD Thesis* Enschede, ISBN 90-9006495-8, 1993:19–23.