

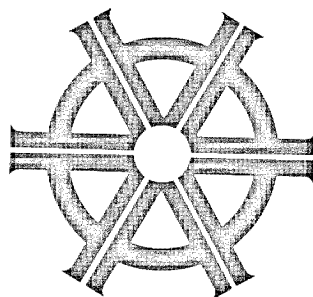
June 1996

Volume 20 • Number 6

Rotary Blood Pumps  
S. Ohtsubo, J.C. Baldwin, and M.E. DeBakey  
*Guest Editors*

# ARTIFICIAL ORGANS

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of the International Society  
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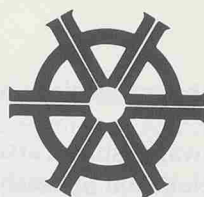
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## Controller for an Axial Flow Blood Pump

\*Hiroaki Konishi, \*†James F. Antaki, \*†Devin V. Amin, †J.R. Boston,  
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\*Mahender Macha, ‡Kenneth C. Butler, \*†Harvey S. Borovetz, and  
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**Abstract:** A rotary blood pump inherently provides only one noninvasive "observable" parameter (motor current) and allows for only one "controllable" parameter (pump speed). To maintain the systemic circulation properly, the pump speed must be controlled to sustain appropriate outlet flows and perfusion pressure while preventing pulmonary damage caused by extremes in preload. Steady-state data were collected at repeated intervals during chronic trials of the Nimbus AxiPump (Nimbus, Inc., Rancho Cordova, California, U.S.A.) in sheep ( $n = 7$ ) and calves ( $n = 12$ ). For each data set, the pump speed was increased at increments of 500 rpm until left ventricular and left atrial emptying was observed by left atrial pressure diminishing to zero. The effect of decreasing

preload was evaluated perioperatively by inferior vena cava occlusion at a constant pump speed. Fourier analysis established a relationship between changes in the pump preload and the power spectra of the pump current waveform. Based on these results, a control method was devised to avoid ventricular collapse and maintain the preload within a physiologic range. The objective of this controller is the minimization of the second and third harmonic of the periodic current waveform. This method is intended to provide a noninvasive regulation of the pump by eliminating the need for extraneous transducers. **Key Words:** Axial flow blood pump—Noninvasive pump regulation.

The axial flow blood pump design has numerous advantages for the next generation of left ventricular assist devices (LVAD): small size, high efficiency, and simplicity of implantation (1). These and other advantages are bought at the expense of inherently greater demand for external control. Unlike their positive displacement counterparts, these devices intrinsically provide only one observable parameter, pump current, and one controllable parameter, pump speed. The characteristic performance of this class of pumps dictates that flow and differential pressure are interdependent. Therefore, it is difficult to control pump flow without flow sensor feedback, or knowledge of afterload or differential pressure (2). To overcome this difficulty, we are attempting to develop a control algorithm based

exclusively on the time-varying electrical power drawn by the pump motor (3,4).

### MATERIALS AND METHODS

An axial flow blood pump (AxiPump; Nimbus, Inc., Rancho Cordova, California, U.S.A.) was implanted in 7 sheep weighing 60–80 kg and in 12 calves weighing 90–110 kg in chronic preparations for 7 to 150 days. The inflow cannula was inserted into the left ventricular apex, and the outflow graft was anastomosed to the descending aorta. Pump flow was obtained by a customized ultrasonic flow probe (Transonic Systems, Inc., Ithaca, New York, U.S.A.) incorporated into the urethane portion of the outflow graft. The pump was placed at the abdominal wall under the muscle. The drive line was connected percutaneously to the drive console, which could vary the pump speed from 6,000 rpm to 13,000 rpm.

Aortic blood pressure (AoP), left atrial pressure

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(LAP), central venous pressure (CVP), pump speed, and pump motor current were collected at 150 Hz on a computer workstation (Apollo, DN3550; Hewlett Packard, Inc. Chelmsford, Massachusetts, U.S.A.). During each data collection, the pump speed was increased at 500 rpm increments from 8,000 rpm to 13,000 rpm. In 3 of these cases, the inferior vena cava (IVC) was occluded gradually before chest closure to decrease the preload of the pump for 15 s.

## RESULTS

Figure 1 shows typical hemodynamic changes observed while the pump speed was increased. Mean AoP increased slightly, and the increase in diastolic pressure was higher than the increase in systolic pressure. Accordingly, the pulse pressure decreased from 40 mm Hg to less than 15 mm Hg. When the pump speed was maximized, collapse of the left ventricle occurred, and the AoP was observed to fluctuate aperiodically. Concurrently, LAP decreased and evidenced an abrupt onset of oscillation as the left atrium was observed to collapse. Continuous or prolonged operation under these conditions, leading to negative LAP and damage to the pulmonary circulation, must be avoided. At the lower extreme, when pump speed was reduced excessively, the flow trace displayed regurgitation from the aorta to the left ventricle during ventricular diastole. Within the preferred operating speed range, the "optimal point", there is no evidence of negative flow.

The shape of the pump motor current waveform was seen to display characteristic changes that were dependent on the pump speed and preload (Fig. 2).

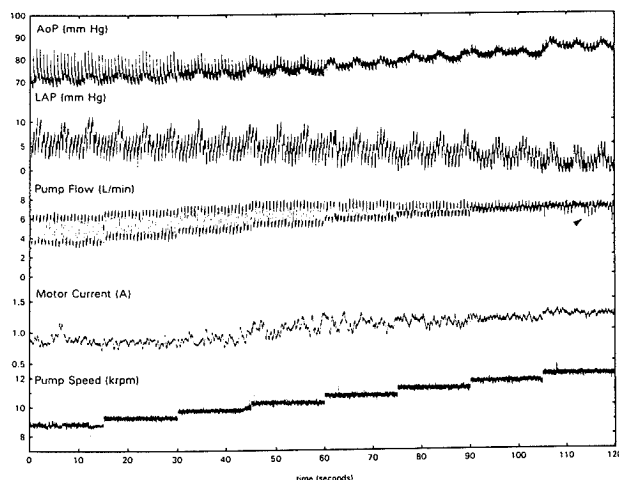


FIG. 1. Hemodynamic changes while the pump speed was increased. AOP, aortic blood pressure; LAP, left atrial pressure. The arrow indicates ventricular collapse.

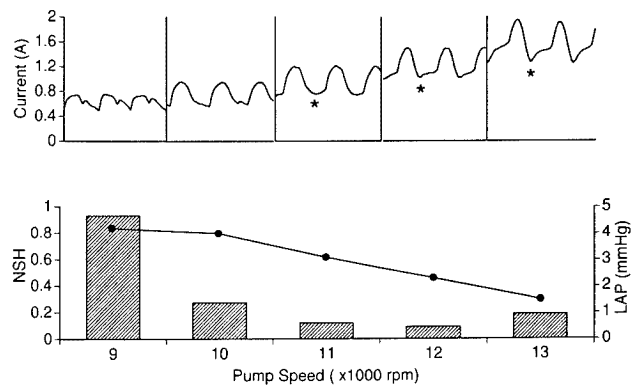


FIG. 2. The pump motor current waveform and the relationship between NSH and LAP are shown. NSH, normalized second harmonic.

A relationship was determined to exist between the optimal pump speed and the normalized second harmonic, defined as the ratio of the second harmonic to the first harmonic (normalized second harmonic [NSH]). The relationship between the NSH and LAP typically revealed a unimodal behavior with a single minimum coinciding with the minimum LAP. This result would imply that if the pump speed is maintained at the point at which the NSH value is minimal, the ventricle and atrium do not collapse and LAP can be maintained within the lower side of its physiologic range.

Figure 3 shows the results during IVC occlusion. During the occluding phase, the LAP and AoP decreased rapidly, and the current waveform became erratic, changing beat by beat. However, during the recovery phase, the preload of the pump returned slowly to physiologic range, thereby displaying the damped response of the current waveform to the changing preload.

## DISCUSSION

In developing an axial flow blood pump as an implantable LVAD, long-term survival has been ob-

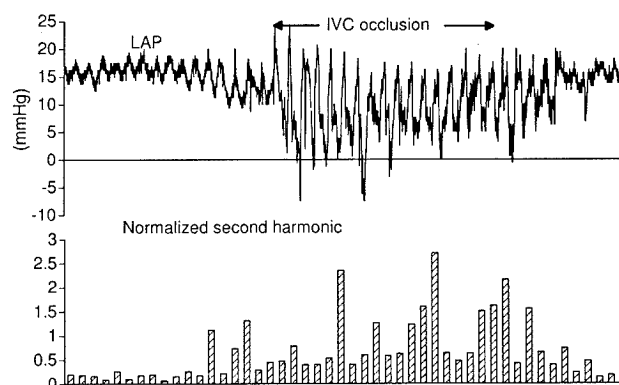


FIG. 3. Results of the inferior vena cava occlusion test are shown. IVC, inferior vena cava.

tained in animals (5). However, there still is no systematic algorithm to control these types of pumps in the chronic setting. Dynamically characterizing the relationship among pump flow, pressure increase, and speed would provide an improved opportunity to control speed for optimal hemodynamics. Although the quasi-static H-Q response of these pumps usually is well defined, the effect of pulsatile hemodynamics introduces significant hysteresis and nonuniqueness (6) (Fig. 4). Accordingly, the ability to identify flow or pressure from speed or motor current becomes much more difficult, if not impossible. In the next-generation controller, to maintain normal hemodynamics under the wide range of conditions expected, pump flow must be adjusted rapidly to accommodate physiologic demand. Because the pump speed is not inherently sensitive to changes in the preload or afterload, the ventricle and atrium will collapse if the pump speed is too high. If the pump speed is too low, regurgitation occurs from the aorta into the left ventricle during ventricular diastole (Fig. 5). Detection of the minimum NSH is one of the noninvasive methods used to avoid these situations. To improve the reliability of the detection of the minimum NSH, it is necessary to obtain a stable motor current signal. In the development of the next stage of this controller, the algorithm used to seek the minimum NSH continuously must be developed. In the second-generation controller, the flow estimation will be derived from the motor current. This work includes

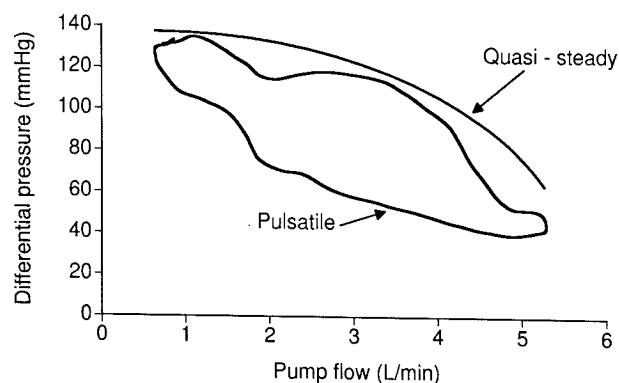


FIG. 4. Pump performance (H-Q curve) (Pump speed = 10 Krpm) is shown.

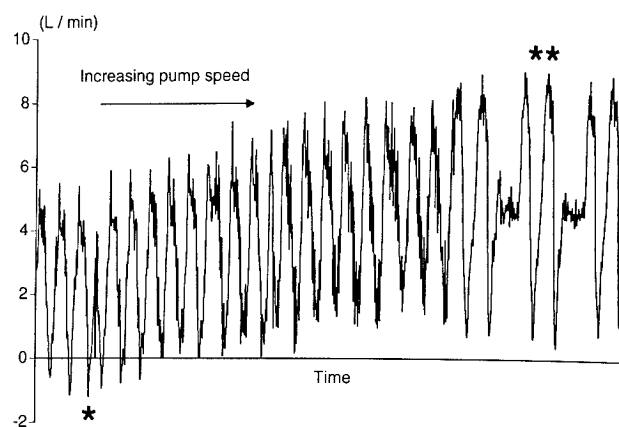


FIG. 5. Pump flow while the pump speed was increased is shown. \*, regurgitation from the aorta into the left ventricle; \*\*, collapse of the left ventricle or atrium.

the application of matched filters to analyze the overall waveform characteristics. Ongoing work to make the control algorithm more robust and enable the control structure to be tuned or adapted to individual patients is another part of this ongoing effort.

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