
Roller Pump Pulsatile Perfusion Utilizing a Constant Stroke Volume Technique

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Introduction

As a result of favorable experience with the intra-aortic balloon pump and extracorporeal balloon pump devices, pulsatile perfusion for clinical cardiac surgery is undergoing a period of increased utilization and investigation. With the introduction of the Cobe-Stöckert roller pump and interfacing pulsatile module, another mode of circulatory support technology can now be applied in the effort to supply an effective substitute for the open-heart patient's native circulation.

In the past, modified roller pumps have been used for pulsatile perfusion, though availability was never widespread.^{1,2,3,4,5,6} The Cobe-Stöckert pump represents a new generation of low inertia roller pumps capable of providing the rapid acceleration and deceleration required to reproduce the rapid, intermittent ejections of the heart. The constant stroke volume technique, designed for use with roller pump pulsatile systems, delivers a consistent and predictable pulsatile circulation, free from the interference of undesirable roller pump artifacts, during total cardiopulmonary bypass.

There have been no previous reports describing a consistent technique for the operation of roller pump driven pulsatile devices. The manufacturers of the

Cobe-Stöckert roller pump system, the only widely available roller pump system with a pulsatile capability, offer no recommended operating technique for the pulsatile operation of the device. The described technique is capable of filling this void.

Materials and Methods

Over 200 adult cardiac patients have undergone pulsatile cardiopulmonary bypass using the described technique. All arterial pressure waveforms were measured from a 20 gauge radial artery catheter via a Gould-Statham P231D pressure transducer and recorded on an 8-channel Hewlett-Packard recorder.

A Cobe-Stöckert roller pump with pulsatile controller module and a nondisposable 24 Fr. (I.D.) stainless steel aortic cannula^a were used in all reported studies. Polyvinyl chloride perfusion tubing, 1/2" I.D., was used in the roller pump chamber in all reported cases with the noted exception of those studies in which 5/8" I.D. silastic tubing was utilized to demonstrate variations in stroke volume and tubing characteristics. An arterial line filter was retained within the arterial circuit since it did not impede pulse generation to a significant degree with this pulsatile system.

As soon as arterial mean blood pressure was maintained with a mean blood flow rate of 6L/m or less (selected arbitrarily as the upper limit for pulsation in

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^a Cardiovascular Instruments, Inc., Wakefield, MA 01880.

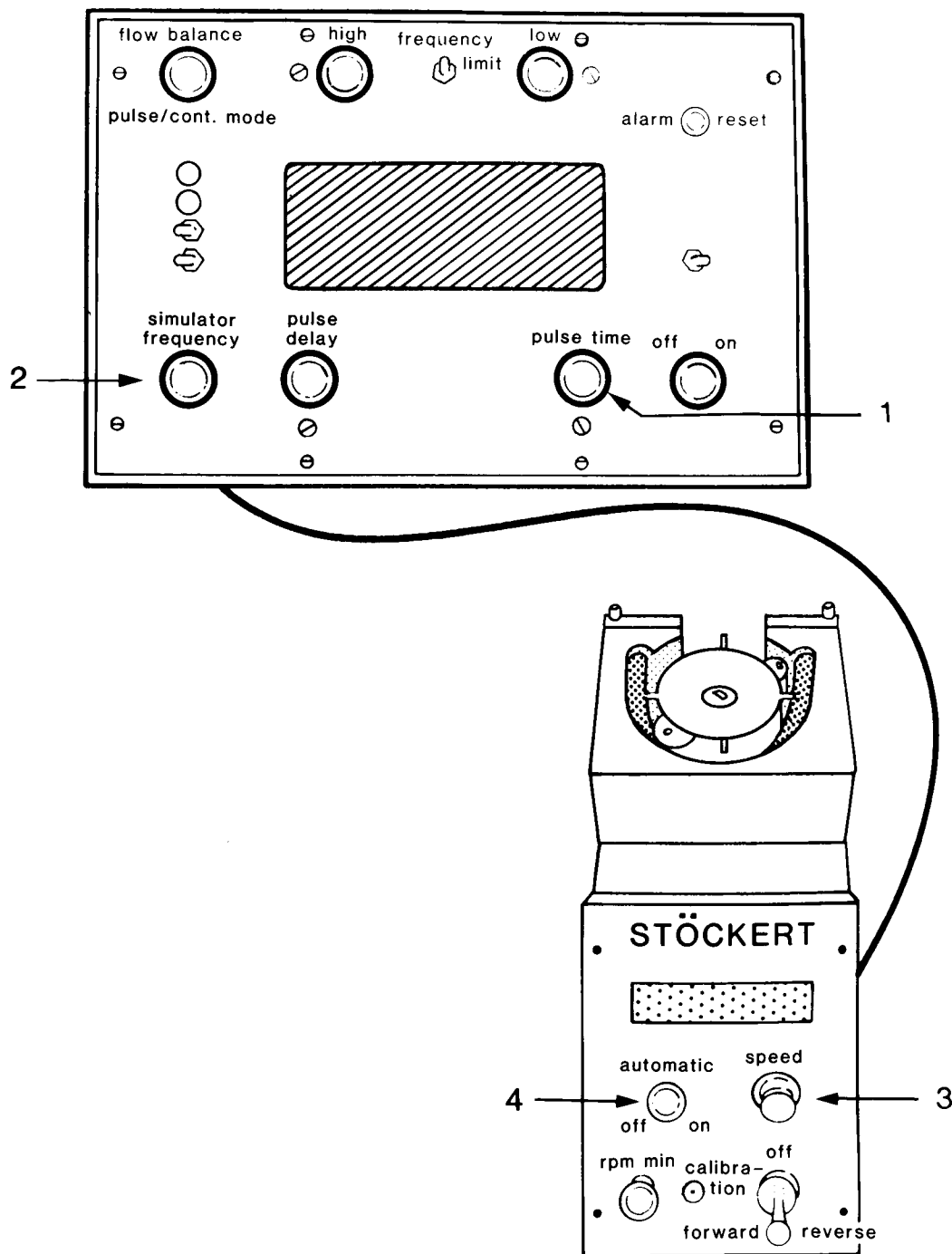


FIGURE 1. Illustration of Cobe-Stöckert roller pump module and pulsatile controller module with interconnecting cable. 1. Pulse Time Control—Controls the length of time the pump will be running during the R-R interval. 2. Simulator Frequency Control—Controls the pulse rate of the pump when the system is being paced by the internal simulator. 3. Speed Control—Governs the rate of rotation of the pump head. 4. Automatic Control—Controls the degree to which the pump speed is influenced by control signals of the pulsatile controller module.

order to avoid unnecessarily high stresses on the arterial circuitry and the perfusionist), and left ventricular ejection was no longer evident, pulsing of the circulation

was commenced. A stroke volume equal to one complete roller pump revolution was fixed (the stroke volume being dependent upon the bore of the tubing in the

arterial roller head) and the required blood flow was attained by adjustments in pulse frequency (blood flow = stroke volume X pulse frequency). By adjusting the stroke volume so that the roller starts and stops at the same point on the roller head, the pulsatile waveform will reproduce itself and can be shaped by altering pulse speed and duration.

With the Cobe-Stöckert pulsatile system, the pulsatile controller module interfaces with a roller pump module by means of an interconnecting cable allowing the pulsatile module to modify the operation of the roller pump module. (see Figure 1) The rate of pressure development is controlled by the speed control (roller pump power control) of the roller pump module and stroke volume is attained by manipulation of the pulse time control of the pulsatile controller module. Pulse frequency is directly set with the simulator frequency control of the pulsatile module. Blood flow is delivered in a completely discontinuous fashion by advancing the roller pump automatic control so that the roller pump comes to a complete stop at the end of each cycle.

The elements of arterial waveform that were analyzed in assessing the adequacy of the technique in effecting a physiologic waveform were ejection time (ET), pulse pressure (PP), and rate of pressure development ($\Delta p/\Delta t$). The ability to maintain the hemodynamic parameters of mean perfusion blood flow rate (\overline{BF}) and the arterial blood pressure (AP) in the face of varying systemic vascular resistance (SVR) while simultaneously maintaining a desirable arterial pressure waveform was the measure of the effectiveness of the technique.

Results

ET during steady state perfusion conditions was usually found to be approximately 30% of pulse duration, though with very high pulse frequencies, this figure may increase to as high as 50%. Though few investigators have studied the significance of this parameter, ejection times of physiologic arterial pressure waves appear to be in the 24%–47% range.^{7,8,9} Pulse pressures of 30–75mm Hg, well within the acceptable 30–40mm Hg range most often reported in the literature,^{10,11,12,13,14} were routinely maintained. Rate of pressure development, measured from the upstroke of the arterial pressure wave, was adjusted to correspond to the patients' own baseline values (approximately 400–600mm Hg/second).

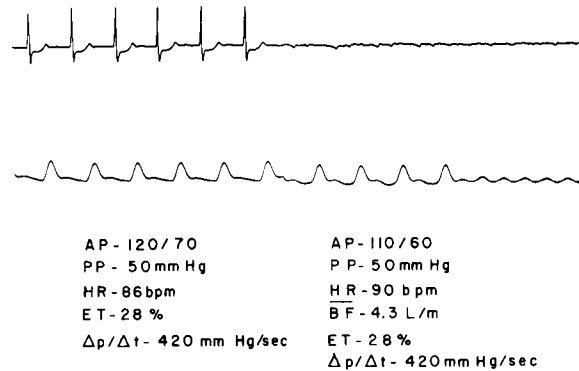


FIGURE 2. Patient baseline arterial pressure recording compared with a good quality pulsatile bypass recording using the described constant stroke volume technique.

Discussion

Arterial pressure waveform is dependent on numerous factors which, for the sake of simplicity, can be summarized as variants of blood flow and/or vascular resistance. Among these factors are stroke volume, heart rate, ejection rate, caliber of the arterial blood vessels, distensibility of the aorta and major arterial blood vessels, and peripheral vascular tone. Other factors such as blood volume, blood composition, autonomic regulation, neuro-endocrine controls, local vasomotor controls, and vasodilator and vasoconstrictor substances also influence the waveform.

The elements that contribute to a graphic representation of the arterial pressure wave are the physiologic forces (blood flow, blood pressure, and SVR) and the artifacts of measurement (arterial catheter size, position, and patency; and the sensitivity or damping characteristics of the electronic monitoring equipment). The influence of variation in the monitoring equipment can be minimized by the comparative analysis of the pre-bypass baseline arterial pressure recordings and the pulsatile waveforms obtained during bypass for each patient. Barring variation in the monitoring system, the physiologic forces and the pulsatile pump characteristics define the quality of the artificial duplication of the arterial pressure waveform. (see Figure 2)

The arterial pressure waveform produced by a pulsatile perfusion circuit is subject to the same factors as a patient's own waveform but is compounded by additional mechanical factors intrinsic to the pulsatile system itself. Two factors in a roller pump pulsatile system are the characteristics of the aortic cannula (i.e. the cannula size, design, and placement) and the caliber

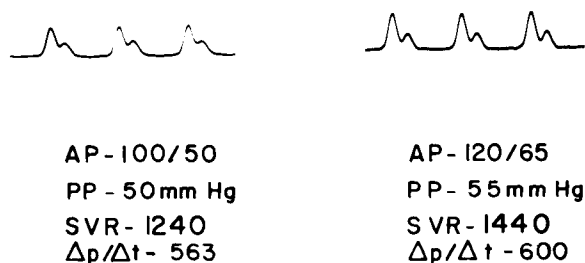


FIGURE 3. Two typical examples of a roller pump pulsatile waveform with "augmented dichrotic notch" tubing artifact.

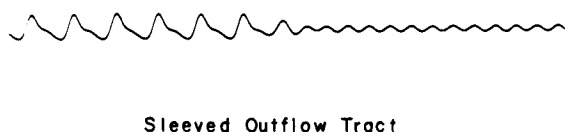


FIGURE 4. Improved waveform and decreased "dichrotic notch" artifact using a sleeved outflow tract.

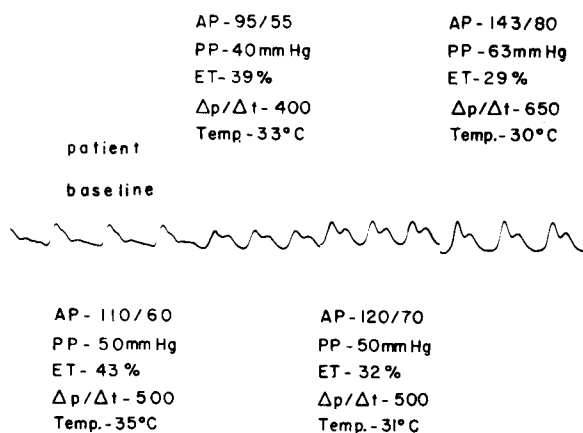


FIGURE 5. A series of arterial waveforms compared with a patient baseline recording shows the increase in pulse pressure and changes in waveform characteristics resulting from the SVR increases associated with hypothermia.

and composition of the cardiopulmonary bypass circuit tubing. A consistent tubing artifact with a roller pump pulsatile system simulating an augmented dichrotic notch (and resembling a counterpulsation wave) results from a small secondary ejection that follows roller cessation and is likely due to a relaxation in the compliant volume built up in the arterial pump circuit tubing during pump systole. (see Figure 3) We have recently decreased the size of this artifact by placing a tubing sleeve around the pump header distal to the

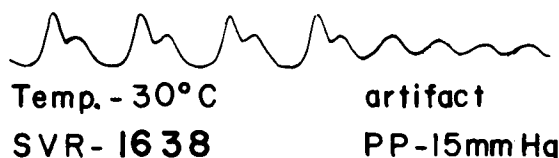
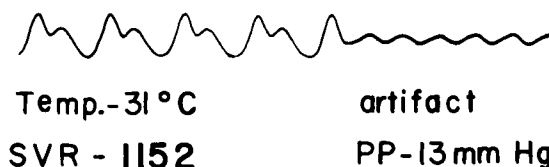
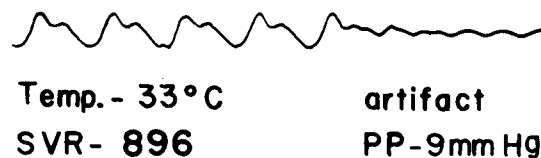


FIGURE 6. The effects of SVR changes on roller pump pulsatile and non-pulsatile waveforms during a hypothermic perfusion.

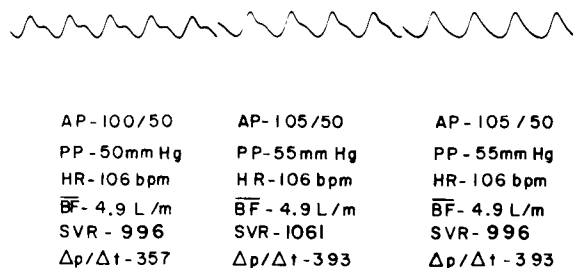


FIGURE 7. Changes in pulsatile waveform due to alterations in pulse speed and pulse duration control settings. Pulse speed is increased and pulse duration is decreased in the figure as the waveform moves in a left to right direction.

rollers to reduce the elasticity of the pump header. (see Figure 4)

During the course of a circulatory support, particularly when hypothermia is utilized, the arterial pressure waveform can be altered by manipulation of the speed control of the roller pump module and the pulse time control of the pulsatile controller module to alter $\Delta p/\Delta t$ and adjust for changes in the systemic vascular resistance. (see Figures 5-7) A constant stroke volume (one revolution/stroke) should always be maintained with the pulse time control, and blood flow rate should always be adjusted by altering pulse frequency. Stroke



AP - 100 / 45
 PP - 55 mm Hg
 HR - 78 bpm
 \overline{BF} - 5.8 L/m
 ET - 30 %
 $\Delta p / \Delta t$ - 375 mm Hg / sec
 PH - 5 / 8 " silastic

FIGURE 8. Pulsatile waveform characteristics with 5/8" silastic pump header. Note the increased stroke volume and decreased pulse frequency (HR).



AP - 165 / 50	AP - 90 / 55
PP - 115 mm Hg	PP - 35 mm Hg
SVR - 1447	
$\Delta p / \Delta t$ - 1375	
PH - 5 / 8 " silastic	

FIGURE 9. Pulsatile waveform with cannula misplacement. Aortic cannula is directed toward the left subclavian artery.

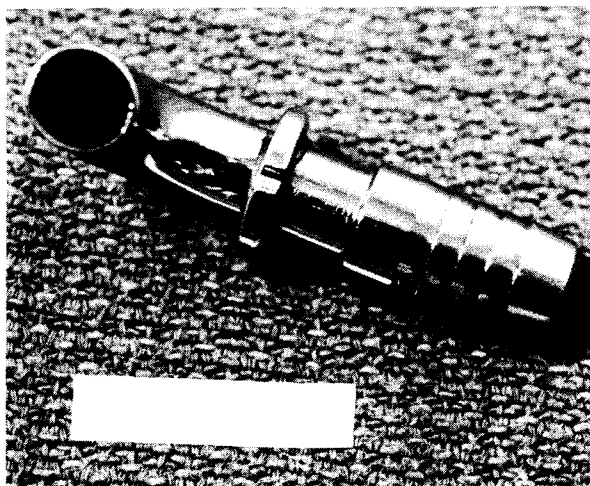


FIGURE 10. Specially designed aortic cannula used in all study cases. (marker = 1 inch in length)

volume can be varied from patient to patient by the selection of from 1/4" I.D. to 5/8" I.D. tubing for the roller head. (see Figure 8)

The importance of the arterial cannula size, design,

BF - 6.0 L/m 24 Fr. CINCO Cannula

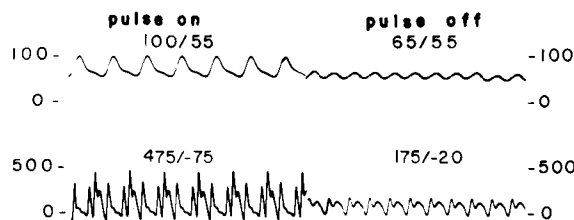


FIGURE 11. Simultaneous pulsatile arterial pressure and perfusion line pressure recordings. (numerical values are in mm Hg)

and placement (see Figure 9), though often not mentioned in the literature, is critical in developing a safe and qualitatively effective pulsatile perfusion system.^{15,16,17,18} After evaluating a plethora of commercially available disposable aortic cannulae, we had a non-disposable, stainless steel, 24 Fr. (I.D.) cannula constructed to our specifications. (see Figure 10) This cannula has proven superior to any of the currently available disposable aortic cannulae and is employed during all of our adult perfusions. Its large bore and thin rigid wall reduce the damping effect on the pulse wave that is frequently reported in pulsatile studies using smaller bore, soft-walled cannulae. The decrease in resistance resulting from the increased bore of the cannula also provides increased protection from shear induced hemolysis.¹⁹ (see Figure 11)

Conclusion

The ease with which blood flow can be readjusted without significant alteration in baroreceptor activity, that is, by altering pulse frequency,²⁰ combined with the elimination of roller artifact made possible by the single revolution pulsing cycles, make the constant stroke volume technique a highly effective and consistent method of roller pump pulsatile perfusion. The ability to effect a pulse pressure, ejection time, and rate of pressure development similar to patient baseline values render this simple technique surprisingly effective. The well established trouble-free clinical operation of roller pumps provides a measure of safety yet to be established with other types of pulsatile pumps and devices.

Despite intensive efforts over nearly three decades, the elements and technology of the perfect substitute

for the native circulation still evades the clinical perfusionist. Currently available roller pump technology, combined with the technique described in this report, provides a more consistent, predictable, and cost-effective method of delivering a pulsatile circulation during open-heart surgery than has previously been described.

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