

Original Article

Significant Safety Advantages Gained with an Improved Pressure-Regulated Blood Pump

J. Patrick Montoya, PhD; Scott I. Merz, PhD; Robert H. Bartlett, MD*

MC3, Inc. and *University of Michigan, Ann Arbor, Michigan

Presented at the 32nd International Conference of the American Society of Extra-Corporeal Technology, Anaheim, California April 8-11, 1994.

Keywords: cardiopulmonary bypass, non-occlusive pump, centrifugal pump, roller pump, safety

ABSTRACT

A prototype of a non-occlusive pressure-regulated blood pump (M-pump) was evaluated *in-vitro* for safety in a comparative study with the roller and centrifugal pumps. The M-pump consists of collapsible tubing of unique design wrapped under tension around a rotor without a stator. The prototype M-pumps were tested *in vitro* to evaluate performance with respect to flow/pressure characteristics, hemolysis, bubble generation (cavitation) and durability. The M-pump and centrifugal pump did not overpressurize at any RPM when the pump outlet was occluded, but the roller pump reached pressures in excess of 1000 mmHg. The M-pump did not generate negative pressures upon occlusion of the inlet, whereas the roller and centrifugal pumps produced near-vacuum pressures. Furthermore, the M-pump was unable to empty a blood reservoir when the height of the pump inlet was placed slightly above the reservoir outlet. The levels of microbubbles in the M-pump were significantly lower than the roller and centrifugal pumps upon sudden restriction of the pump inlet as determined with an ultrasonic bubble detector. The results of our *in-vitro* evaluation of the M-pump have shown it to have lower hemolysis than the centrifugal pump and lower or comparable hemolysis to roller pumps at flowrates of 0.1, 0.5, 4.0 and 6 L/min. We determined that the M-pump design possesses important safety advantages over roller and centrifugal pumps for cardiopulmonary bypass applications.

Address correspondence to:

J. Patrick Montoya, PhD
MC3, Inc.
245 Jackson Industrial Drive
Suite J
Ann Arbor, MI 48103

INTRODUCTION

Roller and centrifugal pumps used in extracorporeal circulation have innate pressure and flow characteristics that make each imperfect for meeting the unique performance demands of blood pumping applications. Limitations of these pumps are typically compensated for with peripheral mechanical and electrical devices. As such, the choice of blood pumps is a personal preference.

In an effort to address many of the pressure regulation safety problems encountered with occlusive roller and centrifugal pumps, we have embarked on improving the non-occlusive peristaltic pump technology originally introduced by Rhone-Poulenc (1). The safety features of the non-occlusive peristaltic pump are many but have not been widely recognized in the United States. We have made significant modifications to the previous technology, and have compared the performance of this new pump concept to the industry's standard roller and centrifugal pumps.

SAFETY PROBLEMS WITH PREVALENT TECHNOLOGY PUMPS

The most commonly used extracorporeal perfusion blood pumps (roller and centrifugal) have adverse pressure regulation properties that could result in fatal accidents in the operating room or intensive care unit. The most recognized hazard with the standard roller pump is that of over-pressurization and circuit rupture if the tubing downstream from the pump (outlet side) is accidentally kinked or blocked. Additionally, if the tubing

upstream from the pump (inlet side) is accidentally kinked or blocked, the roller pump will generate dangerously low negative pressures that could result in blood cavitation, hemolysis, and suction of room air through loose stopcocks or tubing connectors.

Positive and negative pressure control in occlusive roller pumps is typically accomplished with additional electronic equipment such as pressure control modules, or with a mechanical bladder that shuts off the pump if dangerous pressures are generated.

Centrifugal pumps have become popular mainly because, unlike roller pumps, they cannot over-pressurize. However, they can generate very large negative pressures if the tubing upstream from the pump is restricted, with similar results as with occlusive roller pumps. Also, since they are non-occlusive they can allow retrograde flow upon power loss or malfunction (2).

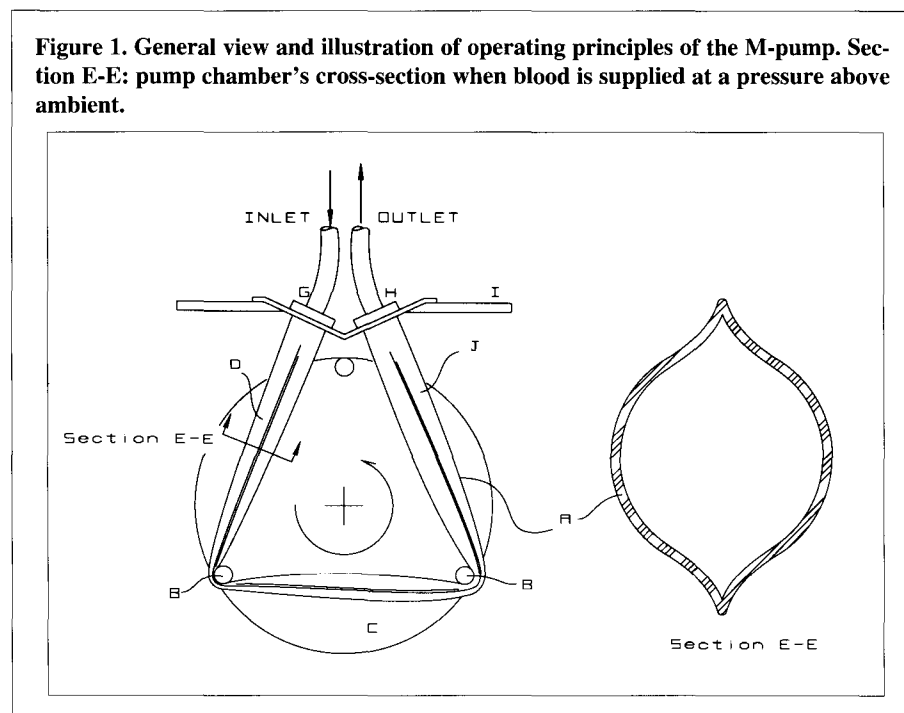
Also, centrifugal and roller pumps are capable of actively sucking blood from an open venous reservoir, and thus are susceptible to emptying the reservoir and pumping air to the patient. It has been claimed that centrifugal pumps hinder passage of air emboli once the pump is deprimed, presumably upon emptying a reservoir (3,4). However, this property should not be perceived as a safety feature or a method of preventing the pumping of air, since it is well known that centrifugal pumps do churn and pump air to various degrees (5).

THE NON-OCCLUSIVE PRESSURE REGULATED PERISTALTIC PUMP

The non-occlusive, pressure regulated (NOPR) pump, also known as the M-pump, is a passively filling, peristaltic pump that eliminates the pressure safety problems associated with roller and centrifugal pumps described above. The basic principles of operation of the pump are illustrated in Figures 1 through 4.

As shown in Figure 1, the pump consists of a flexible tubing (A, the "pump chamber") of unique design, wrapped under tension around rollers (B). The rollers are mounted on a rotatable rotor (C). As the rotor rotates, a peristaltic motion is imparted on the fluid within the pump chamber; when the inlet region of the pump chamber (D) is supplied with blood at a pressure above ambient, the blood is driven by peristaltic motion to the outlet (J). When the pump chamber is supplied with blood at a pressure above ambient, the pump chamber fills, and the cross section assumes the shape shown in Figure 1 (Section E-E).

The pump chamber is completely flat in the portion engaged by the rollers when the pressure within the chamber is equal to or



less than ambient, or when the pump chamber is void of blood (i.e., free, pre-priming shape). Thus, in its free shape, the pump chamber has a cross-section as shown in Figure 2. As shown in Figures 1 and 2, the pump's interior radius of curvature approaches zero at the edges. If the tubing upstream from the pump (pump inlet) is kinked or blocked, the pump chamber will collapse completely into its natural flat shape (Figure 2). When the pump chamber flattens, no further pumping can take place even while the rotor rotates, since there is no blood in the pump chamber available for pumping. The pump chamber lacks the tendency

to return to the inflated condition since it has no memory. Thus, the pump cannot generate negative pressures and will not tend to damage blood and tissue or to suck air through loose tubing connectors upstream from the pump.

This feature can be used advantageously to prevent a hard shell venous reservoir from emptying, since the pump can act as a controller of flow dependent on the available blood level in the reservoir if the pump and reservoir are oriented as shown in Figure 3. The pump inlet is placed at a height even with the minimum allowable level of blood in the reservoir (safety level). As

the level of blood in the venous reservoir approaches the safety level, the pump filling pressure approaches zero, and the pump automatically reduces its flow, completely stopping the pumping of fluid once the safety level is reached. This is the result of the reduction in the pump's filling pressure produced by the pressure of the blood above the pump; as the filling pressure drops, the pump chamber fills less with blood, and eventually stops filling and flattens. Thus the reservoir will never empty even as the pump continues to rotate.

Note that, unlike standard roller pumps, the stator has been eliminated to make the pump non-occlusive (see Figure 1). Thus the maximum outlet pressure the pump will yield is controllable by adjusting the tension (stretching) of the chamber around the rollers. The tension of the chamber is varied by adjusting the distance of the support (I), which holds the inlet (G) and outlet (H) ends of the chamber, further or closer to the rotor (C) (Figure 1). At higher tensions, the chamber becomes increasingly "occluded" at the roller locations, which results in higher pressures being held within the pump chamber. The pump cannot pressurize to a magnitude higher than that controlled by the tension, since the blood would tend to

Figure 2: Pump chamber's cross-section in a free, pre-priming condition. This cross-section is also assumed when blood is not supplied at a pressure above ambient (section F-F). The pump chamber inlet collapses when blood is not supplied at a pressure above ambient.

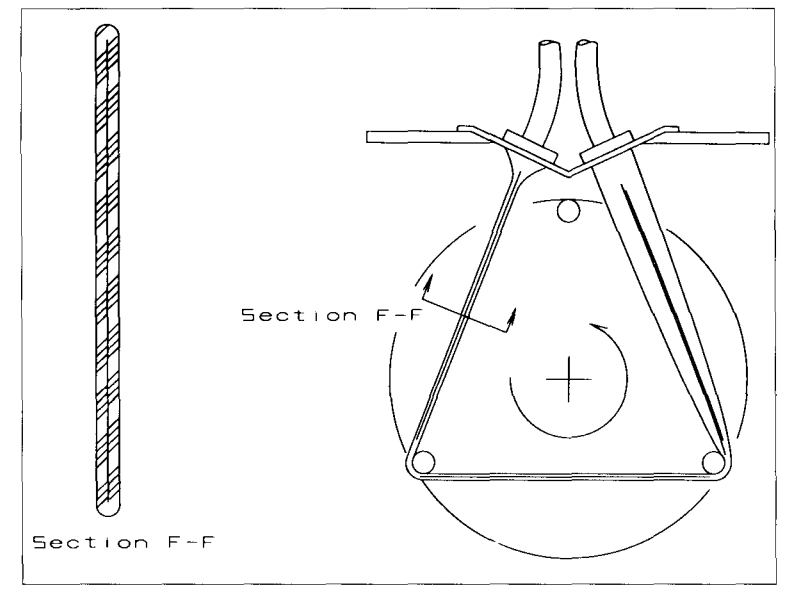


Figure 3: Method of preventing a venous reservoir from emptying by orienting the pump such that the "safety level" in the reservoir is at a height even with the pump inlet.

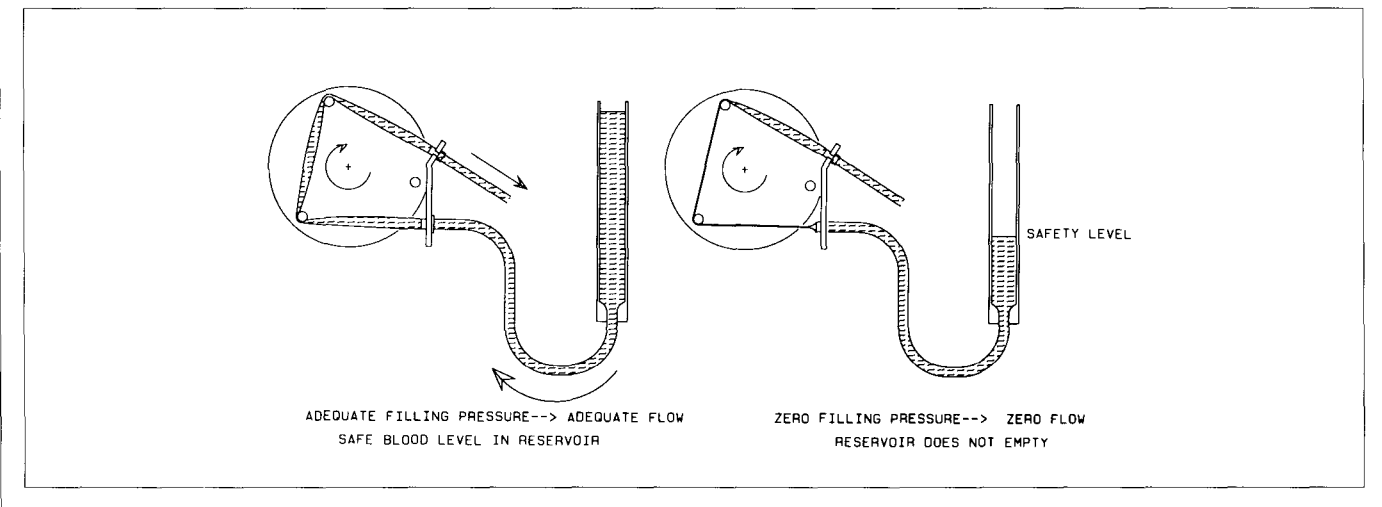
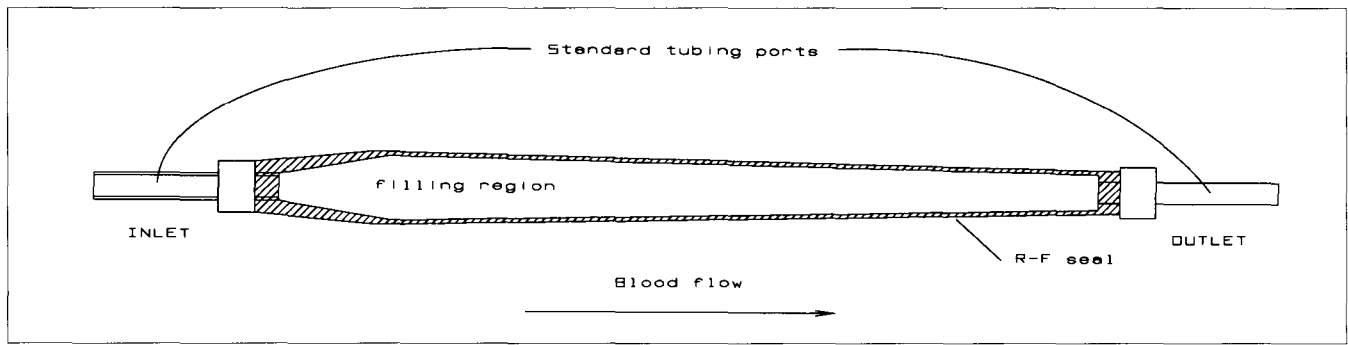


Figure 4: Pump chamber perimeter shape allows for decreasing the flow cross-sectional area from inlet toward outlet to reduce priming volume.



“slip” past the rollers when the pressure exceeded the level retained by the chamber’s tension against the rollers. Hence, the circuit cannot blow-out (as with occlusive roller pumps) when the pump outlet is occluded.

The shape of the chamber allows it to completely flatten at the roller locations without inducing high cyclical bending stresses along the edges of the chamber, as occurs with round tubing in occlusive roller pumps. The design presumably minimizes wear and spallation since the mechanism of repeated bending and compressing of thick round tubing is mitigated (6).

The flow cross-sectional area of the pump chamber decreases from inlet toward outlet (Figure 4) to maximize output flow while minimizing priming volume. A wide filling region just past the pump inlet is required to allow sufficient pump filling and sufficient flow, whereas a wide region toward the outlet would not contribute to increased flow, but would increase priming volume. This is because the outlet region of the M-pump performs more like a conduit; hence, the volume of blood initially trapped between two rollers at the pump inlet will be pumped through the outlet region regardless of the size of the outlet.

The pump chamber is made by radio frequency (RF) sealing two films of polyurethane around the perimeter edges, and RF sealing two standard bypass tubing ports at the ends (Figure 4). This procedure intrinsically provides the desired “free” cross-sectional shape shown in Figure 2. This technique is inexpensive and reliable, and it is frequently used to manufacture venous reservoir bags.

MATERIALS AND METHODS

Two pump sizes were prototyped, a small pump with a 5" rotor and a large pump with a 7" rotor as shown in Figure 5. The corresponding pump chambers were prototyped to perform within the flow ranges indicated in Table 1.

Figure 5: Seven inch rotor M-pump with appropriate pump chamber mounted and primed. The pump is mounted on a mast to allow easy height adjustments. Note the tension adjustment knob on the lower left side.

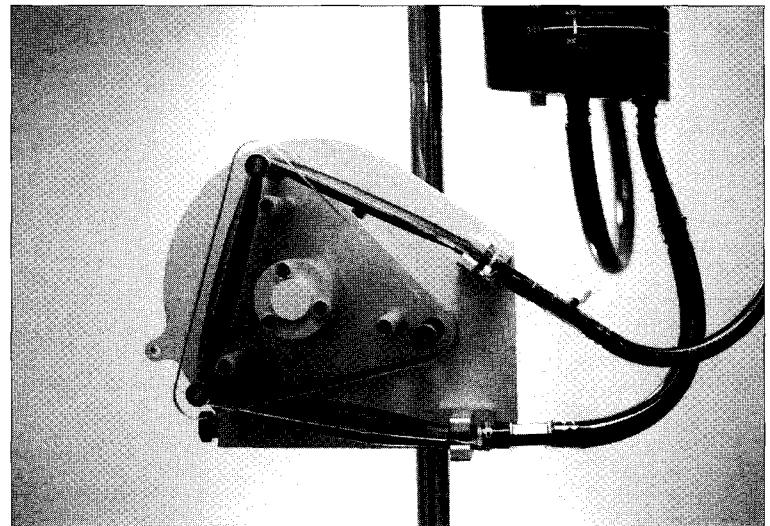


Table 1: Flow ranges and priming volumes for both pump prototypes

size	flow range (ml/min)	priming volume (ml)
small	100-1000	32
large	1000-7000	105

The larger pump chamber is pictured in Figure 6. The rotors were driven with variable speed DC Gearmotors (1/17 and 1/12 HP respectively), which allow for speed control between 3 and 90 rpm. The prototype M-pumps were tested *in-vitro* to evaluate performance with respect to flow/pressure characteristics, hemolysis, bubble generation (cavitation) and durability, according to the following protocols:

Flow-Pressure Performance Characteristics.

The M-pump flow-pressure characteristics were evaluated *in-vitro* using bovine blood at 37°C. A circuit consisting of the M-pump, a blood heater, a venous reservoir bag, and interconnecting tubing was assembled. A pressure transducer was connected to the pump outlet and a clamp-on ultrasonic flowmeter^a was used to measure flowrate downstream from the pump. The reservoir bag was placed 25 cm above the inlet of the pump to allow sufficient pump filling pressure. The circuit was primed with heparinized bovine blood and was allowed to warm to 37°C. A screw clamp was used to variably restrict the pump outlet, and the flowrate and pressure were recorded at various pump speeds and various outlet restriction settings.

Pressure Safety. The pressure safety features of the roller^b, centrifugal^c, and M-pump were evaluated *in vitro* using saline solution. Three identical circuits consisting of a pump, a heater, an open venous reservoir, and interconnecting tubing were assembled. Pressure transducers were placed at the pump inlets and outlets, and a clamp-on ultrasonic flowmeter was used to measure flowrate downstream from the pump. The reservoir outlets were placed 2 cm below the pump inlets. The circuits were primed with saline and were allowed to warm to 37°C. All pumps were set at a flowrate of 6 L/min with a back pressure of 500 mmHg. A tubing clamp was used to occlude the pump inlets or outlets and the flowrates and pressures were recorded. While the pumps were running, the reservoirs were allowed to empty by draining saline through a circuit stopcock to observe the ability of each pump to empty the reservoir and pump air.

Hemolysis. A comparative hemolysis study was performed between the M-pump, the roller pump, and the centrifugal pump at 500, 1000, 4000, and 6000 ml/min. The appropriate pump size was used depending on the test flowrate as indicated in Table 2.

For each test flowrate, three identical circuits consisting of a venous reservoir bag, a blood heater, and PVC tubing were constructed, each with a different pump. Each circuit was primed with 500 - 1000 ml of filtered, heparinized bovine blood from the same batch of blood. Blood was pumped through the circuits at the desired flowrate with a back pressure of 300 or 500 mmHg, which was controlled with an adjustable flow restrictor clamp. The temperature was maintained at 37°C. Samples of

Figure 6: Large size pump chambers side and frontal view

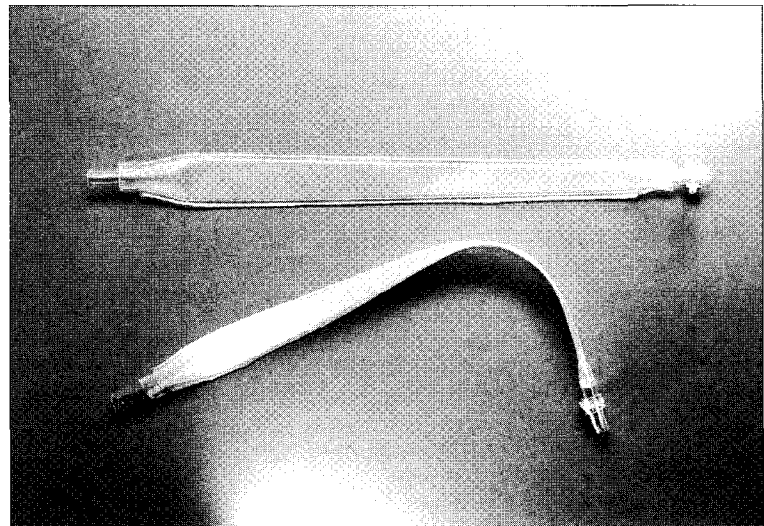


Table 2: Flowrate vs. appropriate pump size

Flowrate (ml/min)	roller pump tubing size (in)	Bio-Medicus BIOPUMP size	M-Pump
100, 500	1/4	BP50	small
4000, 6000	1/2	BP80	large

blood were drawn every 1/2 hour, and plasma was separated and measured for free hemoglobin. The hemolysis index (mg/dl/hr) was calculated for each pump/flowrate combination. The tension on the M-pump was adjusted to a "normal setting" prior to priming, as shown in Figure 3, and the pump tension was adjusted until the chamber barely slacked around the rollers. Note that the roller exerts maximum tension on the pump chamber in the position illustrated in Figure 3, yet by barely slacking the chamber in this position, the tension in all other rotor positions is more relaxed. In earlier experiments it was found that this tension setting produced the lowest hemolysis compared to higher or lower settings.

Bubble Generation. The tendency of the M-pump to cause cavitation and bubble formation of dissolved gases was compared to the roller and centrifugal pumps. Bovine blood with approximately venous blood gas concentrations at 37°C was circulated at 6 L/min in closed circuits using each pump. The inlet to each pump was suddenly clamped for 3 seconds, then released. An ultrasonic flow probe was placed around the tubing downstream from the pump to detect bubbles. The magnitude of the ultrasonic signal transmitted through the fluid path was recorded continuously. The amount of gas extracted from solution and

a Transonic Systems, Ithaca, NY 14850
 b Stöckert-Sorin, Irvine, CA 92714
 c Medtronic Biomedicus, Eden Prairie, MN 55344

Figure 7: Flow/pressure performance characteristics for the large M-pump.

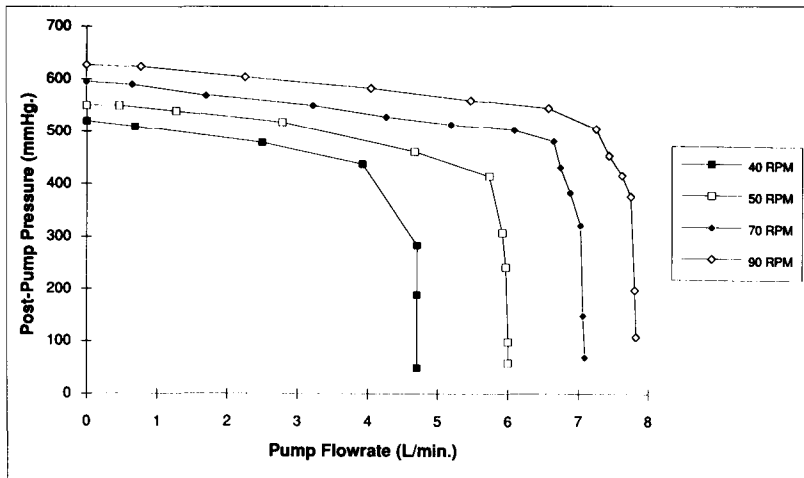
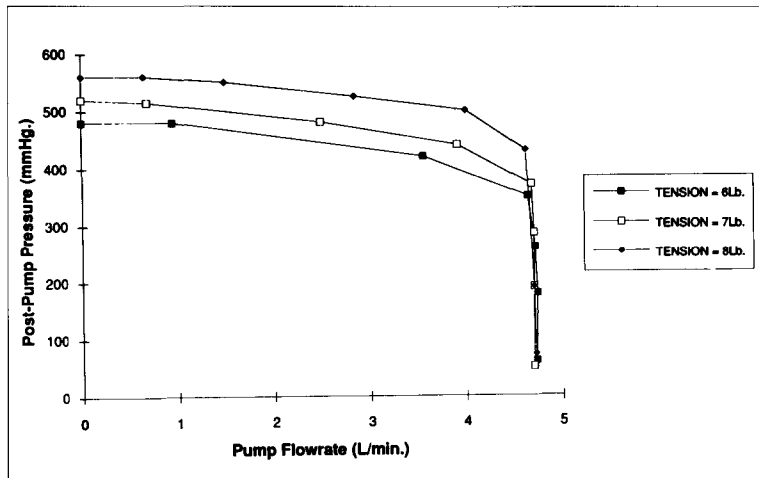


Figure 8: Effect of tension on output pressure.



pumped through the pumps was quantified by measuring the total decrease in ultrasonic signal strength transmitted through the path of the fluid.

Durability. The durability of the M-pump was tested by continuously circulating a 37°C, 60/40% saline/glycerin solution at 6 L/min with an outlet pressure of 500 mmHg. The test circuit consisted of a venous reservoir bag, a large pump chamber, heat exchanger, and interconnecting PVC tubing. The test was performed for 48 hours. Following this, the pump chamber was inspected for cracks or visible wear, then pressurized with air to 1500 mmHg to detect weak spots or impending leaks. Particle spallation was not measured in this study.

Retrograde Flow. The retrograde flow was tested on the M-pump by occluding the tubing downstream from a stopcock immediately at the pump outlet and pressurizing this region to 100,

200, and 300 mmHg with a syringe filled with saline. The tension was adjusted to the normal setting as described above, and the retrograde flow was measured with the ultrasonic flowmeter. The flow was measured with the rotor positioned at maximum and minimum (60° apart from maximum) tension.

RESULTS

Flow-Pressure Characteristics. Figure 7 illustrates the typical performance curves for the M-pump with available filling pressure of 25 cm of blood (HCT=37). The pump's flow-pressure characteristics are similar to those of centrifugal pumps in the low flow/high pressure region, and similar to those of roller pumps in the high flow/low pressure region; centrifugal pumps

tend to maintain a nearly constant pressure over the low flow/high pressure range, whereas roller pumps tend to maintain flow despite the pressure. These performance curves are best interpreted by recognizing that the pump pressure/flow operating point must match those required by the perfusion circuit. For example, if 400 mmHg is necessary to perfuse blood through a circuit and patient at 4 L/min, then these pressure/flow conditions can be achieved at a pump speed of 40 RPM, since this performance curve intersects this pressure/flow point. Similarly, if 500 mmHg is necessary to perfuse blood at 6 L/min through the same circuit and patient, this can be achieved at 70 RPM.

As shown in Figure 7, at zero flowrate (outlet fully occluded) and any given pump speed (RPM), the pressure reaches a maximum value that is within reasonably safe levels. This maximum value is ad-

justable higher or lower, depending on the tubing tension as illustrated in Figure 8 for 45 RPM. At higher tension settings, the performance curve is shifted upward, yielding higher pressures at all flowrates. On the other hand, at lower tension settings the pressures at all flowrates are reduced, yet the maximum flowrates at any RPM remain unaffected by tension.

Pressure Safety. Upon occluding the inlet of the roller pump, the inlet pressure dropped rapidly below -630 mmHg and the saline between the pump and clamp cavitated while the tubing collapsed flat. The centrifugal pump generated negative pressures of -470 mmHg when the inlet was occluded. The M-pump's inlet pressure equilibrated at 0 mmHg (ambient pressure) and generated no negative pressures when the inlet was occluded.

Upon occluding the outlet of the roller pump, the pressure rapidly increased beyond 1500 mmHg (the maximum range of

the pressure transducer) and the tubing burst loose at the connector. The centrifugal pump reached a maximum pressure of 585 mmHg, and the M-pump 570 mmHg when their outlets were occluded.

Upon draining saline from the circuits, the roller and centrifugal pump completely emptied the reservoirs and the tube leading to the pumps, whereas the M-pump did not allow the reservoir liquid level to fall below the outlet, even as the pump rotor continued to rotate.

Hemolysis. The hemolysis indexes for each pump at each flowrate are presented in Table 3. The hemolysis index is used for comparison purposes only within the same experiment (single

flow/pressure), since the recirculation times are different, and fragility of the blood may vary between blood batches. The hemolysis index represents a measure of increase in hemolysis with time, and was calculated from the average of the change in plasma free hemoglobin (mg/dl) divided by the time interval between samples (30 min).

Bubble Generation. The ultrasonic probe used in detecting bubble generation was calibrated by the manufacturer in terms of relating the size of bubbles present to a decrease in ultrasonic signal transmission. A 10% decrease in signal strength results when gas fills approximately 10% of the tubing lumen viewed by the flow probe. Gas filling 50% of the tubing lumen results in a 90% decrease in signal strength.

Figure 9 shows the results of the bubble detection experiment. The horizontal axis represents time in seconds, and the vertical axis represents ultrasonic signal strength, normalized to the baseline value. The pump inlets were obstructed at time = 0 seconds, and released at time = 3 seconds. The roller pump showed the most bubble generation, as ultrasonic signal strength decreased by 100%. The centrifugal pump caused sufficient cavitation to decrease signal strength by 40%. The M-pump caused no appreciable decrease in ultrasonic signal. The experiment demonstrated a potential danger with roller and centrifugal pumps. The fact that the M-pump will not generate negative pressures at any flowrate suggests that bubble generation would not be a problem with that design.

Durability. The M-pump's integrity was unchanged after 48 hours of continuous use. The width of the perimeter seals remained constant throughout the experiment. No weak spots or imminent leaks were detected when the chamber was pressurized.

Retrograde Flow. The M-pump's retrograde flow was measured at below 10 ml/min at pressures of 100, 200, and 300 mmHg. The minimum detectable flowrate on the ultrasonic flowmeter was 10 ml/min.

DISCUSSION

Safety limitations in the performance of roller and centrifugal pumps have been compensated for by the addition of peripheral control devices. We investigated an alternative solution by comparing the safety characteristics of roller and centrifugal pumps to a pump design that attempts to encompass the advantages of both pump technologies. The non-occlusive pump described here, like centrifugal pumps, cannot

Table 3: Hemolysis indexes* for each pump vs. flowrate

Flow @ pressure ml/min @ mmHg	M-Pump (mg/dl/hr)	Roller (mg/dl/hr)	Bio-Medicus (mg/dl/hr)
100 @ 500	11.22 ± 1.06	20.0 ± 1.16	36.53 ± 1.71
500 @ 500	26.57 ± 1.13	26.49 ± 0.51	55.51 ± 0.87
4000 @ 300	15.62 ± 4.90	24.57 ± 5.07	47.34 ± 3.60
6000 @ 500	37.79 ± 2.99	22.78 ± 7.45	104.90 ± 1.83

*Values are mean ± standard error of the mean

Figure 9: Normalized ultrasonic signal strength, associated to the presence of micro gas emboli, plotted against time following sudden occlusion of the inlet to the roller, centrifugal, and the non-occlusive pressure regulated (NOPR) M-pump

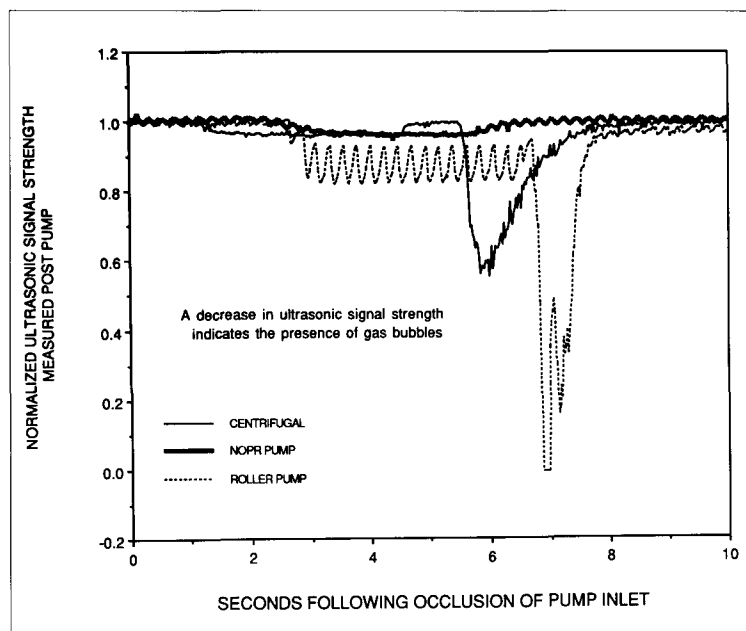


Table 4: Property comparison among studied blood pumps

pump property	roller	centrifugal	M-pump
over-pressurization	yes	no	no
negative pressure generation	yes	yes	no
empty reservoir --> pump air	yes	yes	no
allow retrograde flow	no	yes	no
hemolysis (low flow/high pressure)	low	high	low
hemolysis (high flow/low pressure)	high	low	low

over-pressurize, thus it reduces the risk of blown circuit connections or vessel injury. On the other hand, like roller pumps, the non-occlusive pump impedes retrograde flow in the arterial line upon power loss or malfunction. Also, it is visually obvious when the pump has ceased to rotate, and is easily hand cranked if required.

Another advantage shared with roller pumps is lower hemolysis, compared to centrifugal pumps, in situations where high arterial line pressures are required (7).

Optimal performance with blood pumps requires knowledge of their operation and recognition of their limitations. The operation of the roller pump is intuitive, and it is easily understood how roller pumps can create dangerous circuit pressures, pump air, and cause a circuit to rupture. Centrifugal pumps are simple to operate, but performance can be compromised under certain conditions. For example, significant resistance to flow in the inflow or outflow paths requires high pump speeds, which lead to increased hemolysis and heat generation. Centrifugal pumps tend to perform well at high flows and low pressures, but when higher pressures are required these pumps must spin at very high RPMs which create greater hemolysis. At these high speeds, negative pressure generation and subsequent cavitation of dissolved gases is a concern. Furthermore, air can be sucked into the pump through a loose connector and passed to the patient, as the air is churned with the blood to form bubbles that may be difficult to see.

Our approach to addressing the safety and performance limitations of current blood pumps was to investigate a pump design that possesses inherent performance boundaries that make the pump unable to produce dangerous positive or negative pressures. These boundaries were established to ensure that the pump was capable of generating sufficient flow/pressure output required for any cardiopulmonary bypass (CPB) situation. The price paid for negative pressure safety is the dependence of the pump on a positive filling pressure, and a minimum filling pressure required to achieve maximum output. Techniques exist that

allow the pump to generate limited suction, hence reducing the minimum filling pressure required.

A comparison of safety characteristics of the three pump technologies is summarized in Table 4.

The M-pump design presented here possesses significant safety advantages over roller and centrifugal pumps. Hemolysis rates of the M-pump are generally comparable to the better of the roller and centrifugal pumps over the flow/pressure range of interest. The flow/pressure characteristics and durability of the M-pump are suitable for CPB applications. Most importantly, the M-pump is safe with respect to pressure generation without the need for

additional safety backup measures.

ACKNOWLEDGEMENTS

This research was funded in part by a grant to Michigan Critical Care Consultants, Inc., from a Small Business Innovation Research Grant from the Department of Health and Human Services, National Institute of Health.

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