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In Vitro Testing of a Novel Blood Pump Designed for Temporary Extracorporeal Support

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Abstract

Extracorporeal blood pumps are used as temporary ventricular assist devices or for extracorporeal membrane oxygenation. The ideal pump would be intrinsically self-regulating, carry no risk of cavitation or excessive inlet suction, be afterload insensitive, and valveless thus reducing thrombogenicity. Currently used technology, including roller, centrifugal, and pneumatic pulsatile pumps, does not meet these requirements. We studied a non-occlusive peristaltic pump (M-Pump) in two mock circulatory loops, and compared the performance to a frequently used centrifugal pump and a modified prototype of the M-Pump (the BioVAD). The simple resistance loop consisted of the investigated pump, a fixed height reservoir at 150 mmHg, and a variable inflow reservoir. The pulsatile circulation utilized a mock patient simulator with adjustable resistance elements connected to a pneumatic pulsatile pump. The M-Pump intrinsically regulated flow with changing preload, was afterload insensitive, and did not cavitate, unlike the centrifugal pump. The BioVAD also demonstrated these features, and could augment output with use of vacuum assistance. A non-occlusive peristaltic pump may be superior for short term extracorporeal circulatory assist by mitigating risks of excessive inlet suction, afterload sensitivity, and thrombosis.

Keywords

ECMO; Ventricular Assist Device; Mechanical circulatory support

Introduction

Temporary mechanical circulatory support is often utilized in the treatment of cardiorespiratory failure, either in the form of extracorporeal ventricular assist devices (VADs)^{1,2} or extracorporeal membrane oxygenation (ECMO)^{3,4}. Centrifugal design blood pumps are increasingly employed for these purposes. Centrifugal pumps use an impeller rotating at high speed to accelerate blood in a radial direction. As flow is generated, a vacuum is created, drawing blood into the pump inlet, obviating the need for gravity based drainage and allowing shorter circuits, reduced prime, surface area, and blood transit time. However, the negative pressure generated can be excessive at high rotational speeds. This negative pressure can produce elevated shear forces causing microcavitation, cellular

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damage and hemolysis. Under extreme conditions, it can result in macrocavitation and catastrophic circuit depriming. Furthermore, centrifugal pumps are afterload sensitive, with reduced flow generated at higher outlet pressures. They operate at fixed rotation speed, and may not adequately respond to increases in inflow volume with augmented flow. There may also be regions of low flow and recirculation around the impeller blades, risking thrombus formation.

The M-Pump (Michigan Critical Care Consultants, Ann Arbor, MI) is a non-occlusive peristaltic type pump originally designed for use in cardiopulmonary bypass. We hypothesize that its inherent design overcomes the limitations of centrifugal pumps and thus could be applied as a short term VAD (days to weeks) or for use in ECMO. Specifically, we sought to demonstrate in a mock circulation that the M-Pump, in comparison to a commercially available centrifugal pump, would 1) avoid inlet suction both during high speed operation and during acute inlet line occlusion, 2) respond intrinsically to changing inlet pressures, and 3) was afterload insensitive. The M-Pump however, has a large footprint and requires gravity siphon for drainage, making its existing design cumbersome for use in the intensive care unit (ICU) setting. We have created a novel modified version of the M-Pump (the BioVAD) with an extruded pump chamber, reduced size and a sealed enclosure for regulated vacuum assisted drainage, and tested this initial prototype in the same mock circulation.

Methods

Mock Circulations

Two mock circulation loops were utilized for testing. The first was a simple circuit consisting of a venous reservoir of varying height, and a patient reservoir fixed at 150 mmHg (Figure 1A). The circuit consisted of de-ionized water at room temperature. Inlet and outlet pressures were measured using fluid coupled strain gauges, and flow was recorded using an ultrasonic flow probe. Pumps were assessed at varying venous reservoir heights, pump speeds, and with inlet and outlet occlusion. The second circulation loop was a modification of a patient simulator using a pneumatically driven artificial heart (Cardiowest TAH, Syncardia Systems Inc., Tucson, AZ) (Figure 1B) and four airtight chamber reservoirs with adjustable fluid levels simulating constant compartment pressures of the right and left atria and the systemic and pulmonary circulations. Valves with adjustable occlusion found between the chamber reservoirs simulated systemic and pulmonary vascular resistances. Four separate pressure transducers measured dynamic pressures in each of the four chambers including systolic, diastolic, and mean arterial pressures. Flow in this pulsatile system was fixed at 4 L/min. Systemic resistance was set to achieve a mean arterial pressure of 60mmHg or 110mmHg in the low and high afterload models respectively. This system was used to assess pump performance in a more physiologic, pulsatile model. The extracorporeal blood pumps were connected to the “left atrium” and the “aorta” of the artificial heart.

Pumps

M-Pump—The M-Pump (Figure 2) consists of a collapsible conduit of unique cross-sectional design called the pump chamber, wrapped under tension around three freely rotating rollers mounted on a rotor. The pump chamber will prime and expand only when fluid is supplied at a pressure above the ambient pressure outside the pump chamber. Once the inlet region of the pump chamber is primed, fluid is driven by peristaltic motion. When the pressure inside the pump chamber equals the pressure outside, the pump chamber is collapsed and negative pressure cannot be generated. Outlet pressure generated by the pump is limited by the tension of the conduit around the rollers. At a characteristic pressure limit,

blood simply slips past the roller, and the pressure cannot increase further. At any degree of pump chamber filling between collapse and full filling, the pump acts in a Starling fashion, automatically regulating flow without changes in pump speed. The M-Pump has a pump diameter of 7 inches and a pump chamber volume of 110 mL.

Centrifugal pump—A Bio-Medicus Model 520D (Bio-Medicus, Minneapolis, MN) extracorporeal centrifugal blood pump was used with a Medtronic Bio-Pump+ pump head (Medtronic, Minneapolis, MN) to represent a commercially available centrifugal pump frequently used for clinical ECMO.

BioVAD—The BioVAD prototype is a modification of the M-Pump with a smaller pump diameter of 4 inches and a pump chamber volume of 65 mL. The pump chamber uses an extruded design, rather than the radiofrequency welded chamber used in the M-Pump. The BioVAD pump assembly is a disposable cartridge, with a pre-tensioned chamber. The cartridge is attached to a significantly down-sized drive motor via magnetic coupling. The housing is sealed to allow regulated vacuum-assisted drainage (Figure 3).

Results

In the simple mock circulation, both the M-Pump and centrifugal pump demonstrated increased flow as a function of inlet pressure at fixed operating speeds (Figure 4). At slow speed (50 RPM), the M-Pump lost fluid responsiveness as the pump chamber maximally distends. At higher speed (100 RPM), the pump chamber continued to function in a Starling fashion, augmenting flow with increased inlet pressure. Outlet pressures with increased flow were limited to less than 350 mmHg in both the centrifugal pump and the M-Pump and reached a plateau prior to maximum flow (Figure 5).

The prototype BioVAD was applied to the simple mock circulation. At a fixed inlet pressure of 37 mmHg, application of vacuum improved flow at each pump speed (Figure 6). A maximum of 74 mmHg of vacuum was applied, analogous to 100 cmH₂O gravity siphon typically used in clinical ECMO. With the vacuum turned off, increases in inlet pressure augmented flow, particularly at higher RPM speeds (Figure 4). In general, flow augmentation was not as pronounced with the BioVAD as with the M-Pump. Outlet pressures remained less than 250 mmHg under all conditions (Figure 5).

In the simple mock circulation with a fixed inlet pressure of 50 mmHg, the inlet tubing was fully clamped and inlet pressures measured (Table 1). Gross cavitation and tubing collapse was seen in the centrifugal pump but not in the M-Pump or BioVAD. In the same conditions, the outlet tubing was clamped and outlet pressures measured (Table 2). Pressures were highest in the BioVAD and lowest in the centrifugal pump.

The pumps were connected to the pulsatile circulation model (Table 3). Two systemic vascular resistances were created by adjusting the valve occlusion separating the systemic and right atrial chambers. While there was little or no difference in flow using the M-Pump or the BioVAD, the elevated afterload reduced flow in the centrifugal pump (Figure 7). Inlet pressures were measured during increases in pump speed, demonstrating progressive suction using the centrifugal pump (Figure 8).

Discussion

Implantable mechanical circulatory support using VADs such as the HeartMate II (Thoratec Corp, Pleasanton, CA) and the HVAD (HeartWare International Inc., Framingham, MA) can support the failing circulation and are used either as bridge to heart

transplantation⁵ or for permanent or “Destination” therapy⁶. These durable VADs are highly engineered to provide continuous support for years in a biocompatible fashion. As such, they are expensive (~\$100,000), the patient care protocols are complex, and the infrastructure requirements are expensive. Currently, implantable VADs are only available in specialized centers and applied to a highly selected population⁷. Their use is not ideal for critically ill and acutely decompensating patients in profound shock. However, temporary mechanical support in the form of ECMO or short term VADs using less complex extracorporeal blood pumps can restore circulation and improve end-organ perfusion. The hemodynamic stability allows the time for proper long term treatment strategy to be decided upon, a paradigm now referred to as “bridge to decision”⁸.

A variety of blood pumps are available and marketed for these purposes. The ideal pump would provide satisfactory flow, have biocompatible features reducing thrombogenicity, would intrinsically regulate blood flow based on the patient’s volume status, be minimally afterload sensitive to maintain flow with higher resistance, and carry no risk of hemolysis, cavitation or circuit rupture. Because of these safety features, trained personnel would not be required immediately at the bedside. In addition, costs should be minimized as the mortality rate in this population is expected to be quite high. Pneumatic pulsatile pumps have many of these features, but are costly and thrombogenic. Unregulated roller type pumps carry a high risk of cavitation with insufficient volume delivery, and outlet occlusion would produce excessive circuit pressures and potentially circuit rupture. Roller pumps are only used in ECMO with a trained perfusionist or other qualified personnel continuously monitoring the system. Centrifugal pumps are afterload sensitive, and operate at a fixed rotational speed, giving them limited capacity to auto-regulate based on volume status. In addition, insufficient venous return or inadvertent inlet clamping hazards cavitation with hemolysis or catastrophic circuit de-priming. Centrifugal pumps have regions of high shear associated with mixing of incoming and recirculating blood around the impeller channels. Additionally, the angle of blood contact with the impeller blades varies with flow and pressure and is seldom “on design,” resulting in regions of low flow that present potential areas of thrombus formation, especially with extended use.

The previously described M-Pump satisfies many of the features of an ideal blood pump for short term circulatory assist^{9,10}. In this *in vitro* study, the M-Pump augmented flow with increases in preload without changes in pump speed, a feature which persisted across all pump speeds. This intrinsic autoregulation allows the pump to respond to the patient’s changing conditions without manual interface of the pump settings. The M-Pump maintained flow in the setting of increased afterload, unlike the centrifugal pump which showed substantial drop off. Because of its unique design, there was no extreme negative pressure during clamping of the inlet line during support, while the centrifugal pump grossly cavitated under similar conditions. The peristaltic nature of the M-Pump assures good washout and avoids areas of stagnation, and thus presents an opportunity for improved biocompatibility.

The M-Pump was originally designed and intended for use in cardiopulmonary bypass. The device has a large footprint and relies upon a gravity siphon for drainage, making it impractical for use outside of the operating room theater. The BioVAD prototype is a modification of the M-Pump, with a seamless lay-flat extruded pump chamber, miniaturized for a more accommodating structure, and with a sealed enclosure to allow vacuum assisted drainage. This would obviate the need for gravity siphon, reducing prime and blood transit time. During this initial BioVAD testing, we demonstrated similar features to the M-Pump including intrinsic volume responsiveness, freedom from cavitation, and afterload insensitivity. The blood contacting surface, constructed from a smooth, biocompatible polymer, has no moving parts and can be easily coated with any number of available

products to improve biocompatibility, if necessary. Maximum flow generated was 4.75 L/min, as this prototype was designed for larger children or smaller adults. Adjustments in the pump chamber volume or rotor diameter will increase flow potential. The volume responsiveness of the BioVAD prototype was substantially less than the M-Pump, likely related to the lower volume pump chamber, and can be improved with future modifications. Unlike the M-Pump, negative pressure was generated in the BioVAD during inlet occlusion. The pump chamber of the BioVAD uses an extruded design, as compared to the radiofrequency welded chamber in the M-Pump. This difference resulted in some retention of fluid in the pump chamber during venous line occlusion, resulting in slightly negative pressure. While this degree of inlet suction is tolerable and unlikely to result in hemolysis or cavitation, improvements in pump chamber design can eliminate this finding.

There are several important limitations of this study. The *in vitro* circuit utilized water at room temperature rather than blood or a blood analogue with similar viscosity. Future studies will assess performance of forthcoming BioVAD prototypes using bovine blood or glycerine/water mixtures. However, the simple circuit demonstrated the advantageous features of the M-Pump platform and that these features were reproduced in the first BioVAD prototype. No simulated model can entirely reproduce the complex physiology of the cardiocirculatory system. While the pulsatile model attempts to expand beyond a simple resistance-based loop, many variables cannot be replicated, including dynamic changes in resistance, compliance, left/right ventricular interaction, and alterations in intrinsic cardiac function. Future *in vivo* studies will attempt to elucidate the pump's interaction in a dynamic physiologic system. Durability and hemolysis testing will be performed on future BioVAD prototypes both *in vitro* and *in vivo*.

Conclusion

The M-Pump, and its modified version, the BioVAD, cannot produce excessive inlet suction, carries no risk of cavitation, intrinsically autoregulates to a patient's volume status, and is afterload insensitive. These features may offer advantages over currently available centrifugal pumps when used for short term circulatory assistance. Further modifications in the BioVAD will optimize its flow characteristics.

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Figure 1.

Figure 2.

Figure 3.

Figure 4.

Figure 5.

Figure 6.

Figure 7.

Figure 8.

Table 1

Device	Speed (RPM)	Inlet Pressure (mmHg)	Cavitation
Centrifugal	2000	-23	No
Centrifugal	3000	-285 *	Yes
M-Pump	50	-8	No
M-Pump	100	-6	No
BioVAD	50	-78	No
BioVAD	100	-103	No

* Gross cavitation and tubing collapse were seen, likely underestimating the negative pressure observed.

Table 2

Device	Speed (RPM)	Outlet Pressure (mmHg)
Centrifugal	2000	133
Centrifugal	3000	463
M-Pump	50	437
M-Pump	100	585
BioVAD	50	540
BioVAD	100	644

Table 3

LOW AFTERLOAD							
	No Pump	M-Pump 50 RPM	M-Pump 100 RPM	BioVAD 50 RPM	BioVAD 100 RPM	Centrifugal 2000	Centrifugal 3000
Systolic Pressure	84	153	157	90	102	141	167
Diastolic Pressure	46	75	83	63	81	61	87
Mean Pressure	63	106	110	74	90	89	115
Inlet Pressure	NA	-3	-10	19	-21	29	-37
Pump flow	0	4.2	4.6	2.4	4.4	3.6	6.0
HIGH AFTERLOAD							
	No Pump	M-Pump 50 RPM	M-Pump 100 RPM	BioVAD 50 RPM	BioVAD 100 RPM	Centrifugal 2000	Centrifugal 3000
Systolic Pressure	139	190	193	140	147	177	199
Diastolic Pressure	96	110	116	112	126	92	117
Mean Pressure	116	143	146	124	135	125	148
Inlet Pressure	NA	-4	-10	21	-20	34	-36
Pump flow	0	4.3	4.6	2.4	4.4	2.2	5.1