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Preload Sensitivity in Cardiac Assist Devices

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Abstract

With implantable cardiac assist devices increasingly proving their effectiveness as therapeutic options for end-stage heart failure, it is important for clinicians to understand the unique physiology of device-assisted circulation. Preload sensitivity as it relates to cardiac assist devices is derived from the Frank-Starling relationship between human ventricular filling pressures and ventricular stroke volume. In this review, we stratify the preload sensitivity of 17 implantable cardiac assist devices relative to the native heart and discuss the effect of preload sensitivity on left ventricular volume unloading, levels of cardiac support, and the future development of continuous-flow total artificial heart technology.

Keywords

Circulatory assist devices (LVAD, RVAD,BVAD,TAH); Circulatory hemodynamics; Device; Heart failure

INTRODUCTION

With growing proofs that implantable cardiac assist devices can be effective therapeutic options for end-stage heart failure patients, it is important for clinicians to understand the unique physiology of device-assisted circulation. In the natural heart, preload, afterload, heart rate, myocardial contractility, and compliance interact to determine cardiac output. Increasing myocardial contractility can be understood in cardiac assist devices as increasing pump speed in continuous-flow (CF) devices or increasing pneumatic or hydraulic drive pressures in pulsatile-flow (PF) devices to increase the velocity at which blood is ejected. Cardiac assist devices and the native ventricle have the same inverse relationship between afterload and flow; however, Salamonsen et al. showed that the average afterload sensitivity of four CF rotary blood pumps (HeartWare HVAD, HeartMate II, DuraHeart, and Incor) was almost three times that of the native ventricle [1]. The purpose of this review is to examine available clinical and preclinical data on cardiac assist device output in the context

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of the Frank-Starling ventricular preload sensitivity relationship and to compare the preload sensitivity of pulsatile- and continuous-flow cardiac assist devices with that of the native ventricle.

The concept of preload sensitivity as it relates to cardiac assist devices originates from the relationship between human ventricular filling pressures and ventricular stroke volume, most commonly referred to as the Frank-Starling curves [2]. The term preload refers to the effect of ventricular venous return on ventricular end-diastolic volume. Increasing preload not only increases end-diastolic ventricular volume and pressures but also produces increased stretching of cardiac myocytes prior to the ejection of blood. Similar to increasing the velocity and distance of a projectile by increasing the stretch of a slingshot, the stretching of myocytes increases the force, velocity, and tension generated by the ventricle walls, allowing the ventricle to eject a greater volume of blood at any given heart rate, afterload, and contractile state. This interaction is best demonstrated by the beat-to-beat ventricular pressure-volume loops during decreasing preload conditions where stroke volume is highest with highest end-diastolic filling pressures and volumes (as in the data from a study in canines shown in Figure 1).

It is important that clinicians understand the analogous preload relationship in blood pumps between filling pressures and output for the increasing number of blood pumps being used to safely and effectively treat end-stage heart failure [3–5]. For the purpose of this discussion, pumps are divided into CF or PF devices and into left ventricular assist devices (LVADs), used for univentricular circulatory support, or total artificial hearts (TAHs), in which the native ventricles are removed at the time of implant. We have also limited this review to implantable CF and PF devices only, excluding extracorporeal and paracorporeal devices. Preload sensitivity are quantified for the cardiac assist devices discussed as the ratio of pump output to pump filling pressures in the units L/min/mm Hg.

CF rotary LVADs are currently the devices of choice in clinical practice for heart assist, but PFTAHs remain the only clinically available devices for heart replacement. It is important to understand and compare the preload sensitivity of PF to CF devices, both current and past. A historical perspective on cardiac assist devices demonstrates how CF and PF cardiac assist devices have traded consensus positions as optimal cardiac support mechanisms. LVAD support has swung from primarily short-term extracorporeal CF devices (such as the BioMedicus centrifugal pump), used primarily for cardiogenic shock patients and short-term bridge to transplantation, to the paracorporeal and implantable PF pusher-plate and sac-type pumps for long-term bridge to transplantation, and now recently back to CF rotary pumps as implantable long-term and even permanent devices.

MATERIAL AND METHODS

Continuous-Flow Devices

Preload sensitivity data for the majority of clinical implantable CFLVAD pumps was derived from previously published preload sensitivity or hydraulic performance data: HeartWare HVAD™ (HeartWare International, Inc., Framingham, MA, USA) [1, 6], HeartMate II (Thoratec, Pleasanton, CA, USA) [1, 7], Jarvik 2000 (Jarvik Heart, New York, NY, USA) [1, 7], MicroMed DeBakey (MicroMed, Houston, TX, USA) [8], DuraHeart I (Terumo, Tokyo, Japan) [1], Incor (Berlin Heart, Berlin, Germany) [1], and Impella 2.5 percutaneous CFLVAD (ABIOMED, Danvers, MA, USA) [9]. We also obtained mock circulatory loop *in vitro* preload sensitivity data as per the methods below for both the CorAide CFLVAD (Arrow International, Inc., Reading, PA) [10] which was developed in our laboratories and reached clinical trials in Europe, and the Cleveland Clinic CFTAH [11–12]. Although the CFTAH along with several other single-piece CFTAH devices is still in

early development and animal testing, no other single-piece CFTAH pump has reached clinical use to date with the only clinical reports of CFTAH cardiac replacement being by use of 2 CFLVAD devices to replace native ventricles [13–14].

Pulsatile-Flow Devices

Pulsatile-flow pump data derived from previously published preload sensitivity or hydraulic performance data included: Penn State/3M PFTAH [15], Cleveland Clinic MagScrew PFTAH [16], Utah-100 (Jarvik) PFTAH [17], the predecessor of the CardioWest PFTAH (SynCardia, Tucson, AZ, USA) [18]and the AbioCor single-piece PFTAH (ABIOMED, Danvers, MA, USA) [19]. With the many variables that affect the preload sensitivity of the active filling diaphragm and sac-type PFLVAD, it was not surprising that we could not find quantitative values in the literature for the preload sensitivity of many of these PF devices. As a result, we also obtained mock circulatory loop *in vitro* preload sensitivity data as per the methods described below for the three most commonly used implantable PFLVADs: HeartMate I Pneumatic, HeartMate I XVE (Thoratec, Pleasanton, CA, USA) [20], and the Novacor PFLVAD (WorldHeart Corp., Salt Lake City, Utah) [17].

For this review, we have further divided the implantable PF devices into two categories that impact their sensitivity to preload. Free-filling devices mimic the filling sensitivity of the native heart: the pump chamber fills against atmospheric pressures either by venting the back or air side of the diaphragm or sac to atmosphere through a percutaneous vent tube or by venting it to an implantable air sac that lies against the lungs that are referenced to atmosphere. These pumps are truly responsive to physiologic preloads, as pump filling is not assisted by the use of suction nor is it inhibited by pressures on the back side of the pump chamber. Such pumps are represented by the Penn State/3M PFTAH [15] and the Cleveland Clinic MagScrew PFTAH [16], both of which underwent extensive successful animal testing but did not reach clinical use. The majority of clinical PF devices fall into the second category with regard to pump filling and are here termed active-suction (or active-filling) devices. These PF devices produce negative pressures relative to atmosphere on the back side of the diaphragm or sac to actively assist filling during at least some portion of the blood pump's filling phase. This active-suction can occur by vacuum applied to the drive line of pneumatic PF devices during the pump-filling phase, as in the Utah-100 (Jarvik) PFTAH [17], the predecessor of the CardioWest PFTAH (SynCardia, Tucson, AZ, USA) [18]. Suction can also be generated as a result of the material modulus and thickness of a pump diaphragm or sac. This type of PF pump creates significant stress in the pump's diaphragm as actuators push the diaphragm into full eject position and then produce significant suction pressures during filling, when the polymer-restoring forces then act over a relatively large diaphragm surface area to pull the diaphragm into a fill position. Active-suction PFLVADs include the HeartMate I Pneumatic and HeartMate I XVE (Thoratec, Pleasanton, CA, USA) [20], and the Novacor PFLVAD (WorldHeart Corp., Salt Lake City, Utah) [17]. The AbioCor single-piece PFTAH (ABIOMED, Danvers, MA, USA) [19] can generate interventricular hydraulic suction, which acts on its dual pump sacs during operation.

In Vitro Mock Circulatory Loop Experimental Methods

The Cleveland Clinic LVAD mock circulatory loop was used to evaluate the CorAide CFLVAD pump and the HeartMate I, HeartMate I XVE PFLVAD and Novacor PFLVAD pumps. The test pump inlet port was connected to an open fluid reservoir. The outlet port was connected to a sealed fluid-/air-filled compliance chamber, and a clamp simulated systemic vascular resistance. The height of the fluid in the open reservoir was recorded as the filling pressure. Starting at a minimum filling pressure of 3 mm Hg, fluid was then incrementally added to the loop to increase filling pressures up to 30 mm Hg. At each

steady-state condition, the pump's flow was recorded after adjustments in systemic resistance were made to keep arterial pressure constant at either 80 mm Hg or 100 mm Hg. To normalize the functional contractility of the CorAide CFLVAD, the pump speed was fixed for all preload conditions at whatever speed produced a flow of approximately 4.0 L/min at the minimum loop filling pressure of 3 mm Hg. PFLVAD pumps were run in a preload-sensitive fill-to-empty mode, in which ejection of blood occurred as soon as the pumps reached a fixed level of filling (approximately 90%). The filling pressure vs. pump flow data was analyzed by linear regression with the slope of this analysis recorded as the pump preload sensitivity (L/min/mm Hg).

The CFTAH mock circulatory loop [11] was used to evaluate the preload sensitivity of the Cleveland Clinic CFTAH at a fixed speed of 2,600 rpm. An illustration and diagram of the *in vitro* test setup is shown in Figure 2. The same working fluid and pump preload levels were tested. At each preload condition, systemic and pulmonary vascular resistance was adjusted to maintain systemic and pulmonary pressures at 100 and 20 mm Hg, respectively.

Human Preload Sensitivity Baseline

Given the complexity of the human preload sensitivity relationship, it is difficult to derive a single preload sensitivity value for humans; however, Salamonsen *et al.* [1] have derived a value of 0.213 ± 0.03 L/min/mm Hg using Guyton's Starling curves and extrapolated human data. This value is supported by a measured preload sensitivity value of 0.275 L/min/mm Hg obtained by Levine *et al.* [21] for saline volume loading in healthy adult males at rest. The average resting cardiac output increased from 4.3 to 7.7 L/min as pulmonary capillary wedge pressure increased from 5.1 to 17.6 mm Hg. Also included with the plotted data is an estimate of the preload sensitivity of a failing human heart based on clinical experience and not experimental data.

RESULTS

Preload Categorization of Cardiac Assist Devices

Preload sensitivity values for all pumps, derived from either direct testing or from previously published data, are listed in Table 1 and illustrated in Figure 3. Our review of the preload data for these pumps yielded the following observations: 1) CF pumps show the lowest average preload sensitivity (0.080 ± 0.038 L/min/mm Hg) at approximately one third that of the human left ventricle at rest (0.241 ± 0.040 L/min/mm Hg) and almost one seventh that for all PF pumps (0.537 ± 0.507 L/min/mm Hg). 2) PFTAHs showed the highest preload sensitivity, averaging 0.860 ± 0.436 L/min/mm Hg for both the free-filling (vent tube or compliance chamber reference pressures) and active-filling suction (pneumatic or hydraulic) TAH devices. 3) Preload sensitivity of the CFTAH (0.102 L/min/mm Hg) was similar to that of CFLVAD pumps (0.077 ± 0.040 L/min/mm Hg). 4) The only pump data not following the trend of high preload sensitivity for PF pumps was the relatively low average preload sensitivity (0.107 ± 0.015 L/min/mm Hg) obtained for PFLVAD pumps with active suction-assisted filling (Novacor, HeartMate I Pneumatic and HeartMate I XVE PFLVADs. This finding is illustrated in Figure 4, which graphically compares the full preload sensitivity data set for the pumps tested in our laboratories vs. that reported for a healthy human at rest and that estimated for a patient with severe heart failure. The active-suction PFLVAD pumps showed very high outputs (6.00 ± 0.36 L/min) at only 3 mm Hg preload with low preload sensitivity for filling pressures above 3 mm Hg. This finding is explained by the dynamics of pump filling for these pumps. Their high filling rate is due primarily to the high suction levels created during the first part of the pump's filling phase, making the filling rate independent of preload when tested *in vitro* with an open venous reservoir. The high filling rates produce high beat rates and flows even at low filling

pressures when operated in fill-to-empty operating modes, in which the pump is ejected only after reaching approximately 90% filling capacity.

COMMENT

These compiled LVAD and TAH data strongly support the generalization that CF rotary blood pumps have lower preload sensitivity than the human heart, whereas free-filling PF pumps have much higher preload sensitivity relative to the heart. In our *in vitro* testing, we found little difference in the preload sensitivity values of pumps tested at afterloads below 100 mm Hg; however, Salamonsen *et al.* reported significantly higher preload sensitivity values in centrifugal CF rotary pumps when afterloads were increased to 140 mm Hg. Evaluation of preload sensitivity at very high afterloads, however, is not clinically relevant as operating any CF pump at high afterloads is strictly contraindicated due to their high afterload sensitivity which will lead to low pump outputs predisposing the devices to thrombus formation, a complication of far greater importance than changing levels of preload sensitivity. It is also not recommended in that all current CF pump require anticoagulation which when combined with high systemic pressures can predispose this patient population to bleeding and hemorrhagic cerebrovascular events.

Impact of Preload Sensitivity on Left Ventricular Unloading

Clinical studies showing that PFLVADs produce more LV volume unloading than CFLVADs may be explained by higher PFLVAD preload sensitivity [22–23]. This volume unloading is generally reported as a greater reduction in LV end-systolic and end-diastolic dimensions and volumes when PFLVADs are used. The findings of Letsou *et al.* [24] in controlled animal studies showed that fill to empty synchronization of a PFLVAD in counter pulsation with the LV produces superior LV unloading compared with CFLVAD at the same flow rates. A report by Garcia *et al.* [25] found no difference in volume unloading between the HeartMate XVE PFLVAD vs. the HeartMate II CFLVAD; however, the authors noted that clinical practice dictated that the PF HeartMate pump was run at a fixed beat rate in order to reduce mechanical wear and not in the preload-sensitive fill-to-empty mode. At a fixed beat rate, PFLVADs are asynchronous with the native cardiac cycle, cycling between pump filling during LV systole and then LV diastole. In fill-to-empty mode, they naturally synchronize with the cardiac cycle in counterpulsation, which greatly improves LV volume unloading because the pumps always fill during ventricular systole.

Klotz *et al.* [22] further clarified LV unloading in a clinical review comparing the MicroMed DeBakey CFLVAD to the Novacor and HeartMate VE PFLVADs. The authors found that CF and PF devices provided similar LV pressure unloading; however, PFLVADs produced pronounced LV volume unloading. In spite of the larger LV volumes seen in CFLVAD support, there has been no significant pathology reported as a result of this factor. Haft *et al.* [23] reported that the HeartMate XVE PFLVAD and the HeartMate II CFLVAD provided equivalent degrees of hemodynamic support and exercise capacity, even though the XVE was associated with significantly greater LV volume unloading. In a recent study by Kato *et al.* [26] of 61 patients evenly divided between CFLVAD and PFLVAD using primarily the HeartMate II and HeartMate XVE, respectively, the PFLVAD demonstrated smaller LV end-diastolic and end-systolic dimensions, significantly better systolic and diastolic function by serial echocardiography and a greater decrease in myocardial and circulating B-type natriuretic peptide. Both groups, however, showed reduced myocardial hypertrophy and myocyte cross-sectional area indicating a reduced work load on the heart. These authors suggest that based on these findings, LV unloading by PF devices may dictate the class of device selected for the bridge to recovery LVAD patient population.

The LV volume unloading issue has not, however, decreased current enthusiasm for CFLVADs over the larger, less reliable valved PFLVADs. This has been primarily driven by the smaller size of the CFLVADs allowing for a larger potential LVAD patient population and less surgical complications, greater reliability demonstrated by 5 to 10 year expected operating life, less foreign body mass and surface areas, smaller blood contacting surfaces with less intra-pump stasis, smaller drive lines and reduced drive line infections and lower power requirements all with low thromboembolic complication rates similar to the PFLVAD devices which preceded them. In a randomized trial comparing outcomes for the HeartMate XVE PFLVAD and the HeartMate II CFLAD in advanced heart failure patients who were ineligible for transplantation, the CFLVAD demonstrated a higher (46% vs. 11%) incidence 2 year survival free of stroke, a higher actuarial survival rate at 2 years (58% vs.24%), a lower incidence of device-related infections, right heart failure, respiratory failure, renal failure, cardiac arrhythmia and need for device replacement [3].

Significance of Preload Sensitivity to Function and Control of Continuous-Flow Biventricular Assist Devices and Total Artificial Hearts

For treatment of severe biventricular failure, several clinical groups have recently reported successful outcomes with biventricular CF circulatory support when the ventricles are left intact [27–30]. The use of biventricular support in TAH applications in which the native ventricles have been removed has been very limited, however. Total heart replacement devices will support a heart failure population subset that cannot be served by biventricular support due to specific cardiac pathology, such as ventricular wall aneurysms, severe valve disease, recurrent ventricular and atrial wall thrombi, intractable ventricular tachyarrhythmias, or severe right and left heart failure, or those patients with myocardial infarction for which ventricular cannulation is difficult or risky. For this application, in which the native ventricles are no longer available to compensate for flow and pressure mismatches, adequate preload sensitivity is critical to ensure right and left atrial pressure and flow balance and to prevent atrial or ventricular remnant “suck down” at the pump inlets. Ideally, a continuous-flow biventricular cardiac replacement system should be able to respond to a 5- to 10-mm Hg increase in atrial pressures resulting from a transient right/left flow imbalance by increasing the output of the underperforming pump. If preload sensitivity is too low, the flow-increase response will be insufficient, leading to problems of pulmonary or systemic congestion. Similarly, if atrial pressures drop transiently or because of chronic factors such as dehydration, ideally a pump will decrease its output to prevent suction effects at the inlet. Free-filling, high-preload-sensitive PFTAHs respond to decreased atrial pressures by decreasing the rate of filling and thus the rate of ejection and output because these TAHs are typically operated in fill-to-empty mode, in which ejection only occurs after pump filling. The high preload sensitivity of PFTAHs has allowed these devices to achieve good flow and atrial pressure balance and is the primary reason they are the only current treatment option for severe biventricular failure. The CF devices have not been able to adequately identify low atrial pressures to prevent suction events or increase output in response to increased venous return, which has historically prevented their application to total heart replacement.

Strueber *et al.* [14] and Loebe *et al.* [13], however, have reported encouraging yet very limited data on their initial clinical experience with total cardiac replacement using dual HeartWare HVADs and dual HeartMate II CFLVADs, respectively. It is believed that for the low-preload-sensitive CF rotary pumps to provide safe and effective total cardiac replacement, more sophisticated biventricular CF control algorithms and/or pressure/flow sensors will be needed. This currently limited yet exciting area of clinical research focuses system and device design and control on providing adequate preload sensitivity to provide

physiologically responsive outputs while preventing suction events. This represents significant engineering challenges that will require major technical advancements.

Growing interest in the development of single-piece CFTAH devices for cardiac replacement has prompted evaluation of the BiVACOR BV Replace [31] and our Cleveland Clinic CFTAH [11–12]. Because of the early developmental stages of these CFTAH systems, it is yet to be determined whether CFTAHs or PFTAHs will be better for the subset of heart failure patients in which the native ventricles are removed. Although it is too early to provide reliable data on average total pump outputs for CF biventricular cardiac replacement, we know that the average pump flow reported for the HeartMate II CFLVAD was 5.6 L/min [32] vs. 6.4 L/min (average cardiac index of 3.2 L/min/m^2 and body surface area of 2.0 m^2) for the CardioWest pneumatic PFTAH [33]. This relatively higher PF output can be attributed to its greater preload sensitivity as well as the larger size of the PF devices. Pump size is less of a problem for TAHs than for LVAD support because TAHs are usually placed in large remodeled pericardial spaces, replacing enlarged diseased hearts. Also, although it has been proven that delivery of essentially resting cardiac outputs by the CFLVADs provides a very acceptable level of perfusion in terms of patient outcomes and quality of life, limiting total output to resting physiologic levels in cardiac replacement therapy may be a disadvantage where residual native ventricular output is not available during periods of increased metabolic demand.

Limitations to interpretation of the data in this review article were some variability in preload sensitivity values reported in the literature, errors inherent in estimating preload sensitivity based on published pump performance curves at various pump speeds as well potential variability in derived values based on different test conditions. Thus, we did not compare individual device preload sensitivity within the CF or PF classes of pumps. This variability, however, did not significantly affect our qualitative comparison between the 2 classes of pumps and the native ventricle given their obvious differences.

Conclusions

The native ventricle operates on a series of Frank-Starling preload curves, each defined by the afterload and contractile state of the heart. Figure 1 is a snapshot of how varying preload end-diastolic and end-systolic LV volumes and pressures affect stroke volume at one fixed afterload and contractile state. Thus, preload sensitivity – independent of afterload and ventricular contractility – is a concept used to understand cardiac physiology, and we use the same principles here to understand the preload sensitivity of cardiac assist devices and the effect of preload sensitivity on cardiovascular physiology.

Consideration of these factors has increasing clinical significance as cardiac assist devices, and CFLVADs in particular, transition from experimental devices controlled by cardiac surgeons to what is now the accepted method for clinical management of patients in end-stage heart failure who do not respond to less invasive treatment options. As the number of patients with implanted devices increases, postoperative management is becoming the province of a growing number of cardiologists who must manage the cardiovascular disease that these patients still have but in a new context – that of cardiovascular physiology under assisted circulation. Cardiologists specializing in heart failure are increasingly referring patients for elective device implantation before the patients reach a critical care stage [34]. Permanent cardiac assist devices are also becoming a choice for patients who want to trade the uncertainty and poor quality of life experienced in end-stage heart failure for the stable hemodynamics and greatly improved quality of life that these devices appear to be providing [35]. The National Heart, Lung, and Blood Institute is conducting a clinical trial to examine the efficacy of a CFLVAD vs. medical therapy for non-inotrope-dependent New York Heart Association (NYHA) class IIIB patients [36].

Conclusions

The stratification of preload sensitivity among the major classes of cardiac assist devices relative to the native heart and the effects of device preload sensitivity on LV volume unloading, levels of cardiac support, and the future development of CFTAH technology have been presented. It is hoped that this review will help clinicians better understand the unique cardiovascular physiology presented by patients with cardiac assist devices, allowing their physicians to better evaluate current and future devices and, most important, improve patient care.

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Abbreviations and Acronyms

CF	continuous-flow
CFLVAD	continuous-flow left ventricular assist device
CFTAH	continuous-flow total artificial heart
LV	left ventricular
LVAD	left ventricular assist device
PF	pulsatile-flow
PFLVAD	pulsatile-flow left ventricular assist device
PFTAH	pulsatile-flow total artificial heart
TAH	total artificial heart

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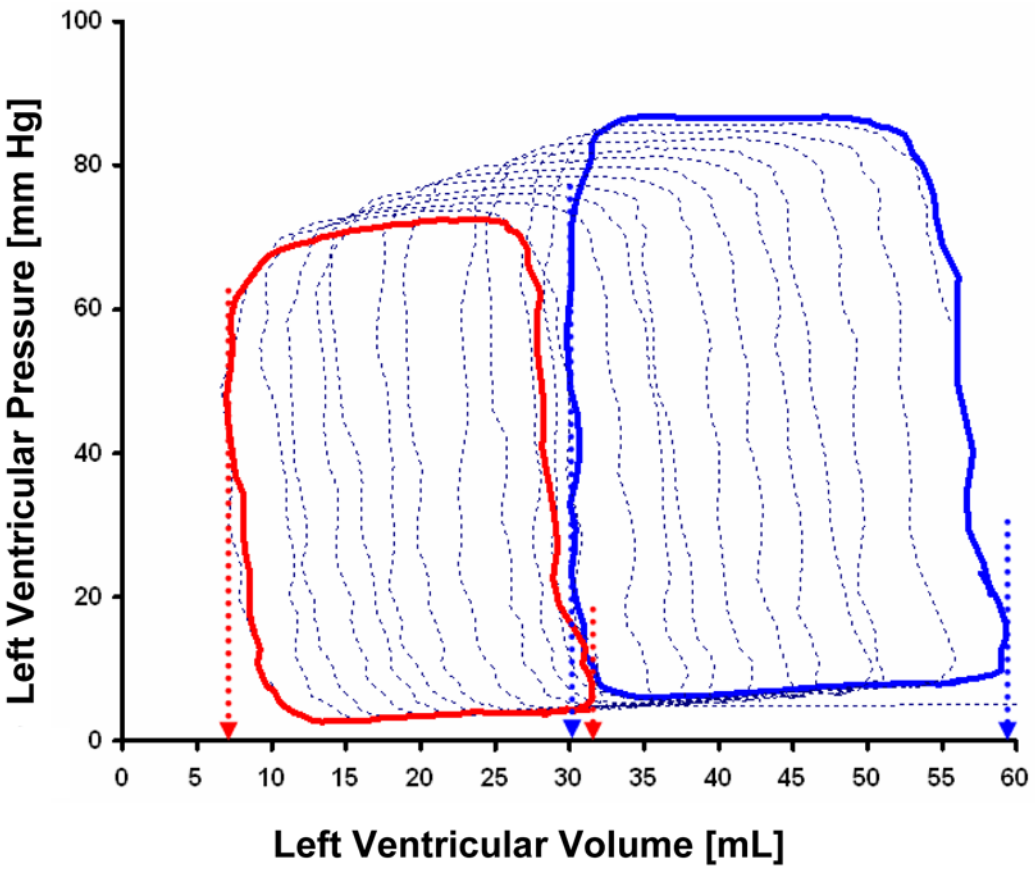

Figure 1.
Typical left ventricular pressure-volume loops during inferior vena cava occlusion obtained from a healthy canine in an open-chest condition. As preload decreases, stroke volume decreases proportionally as the decrease in end-diastolic volume exceeds the decrease in end-systolic volume. There is a 20% decrease in stroke volume from 30 ml (blue loop) to 25 ml (red loop) with decreasing preload.

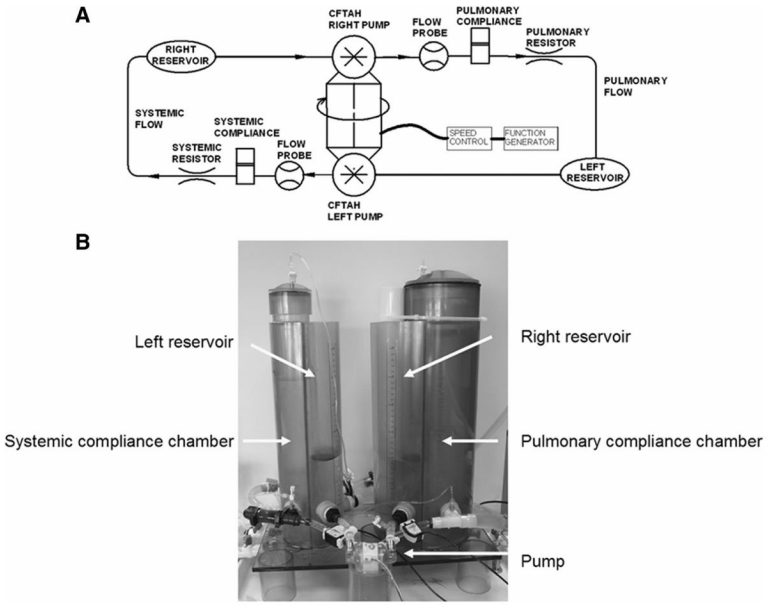

Figure 2.
Schematic drawing (a) and picture (b) of the total artificial heart mock circulatory loop.

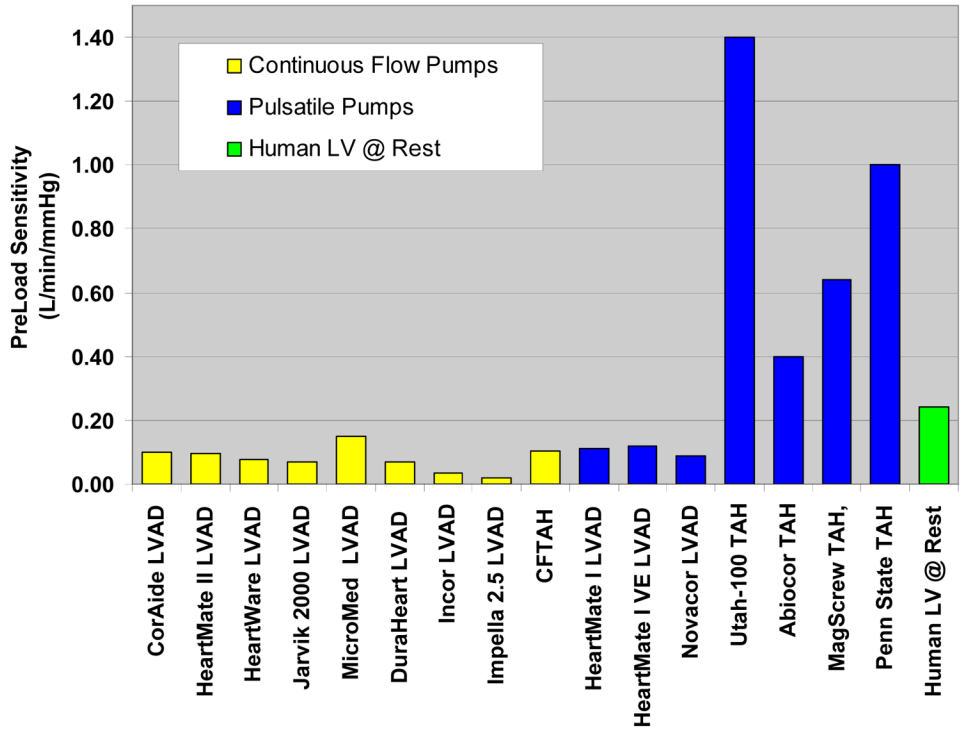

Figure 3.
A comparison of preload sensitivity of continuous-flow and pulsatile-flow devices with that of human left ventricle.

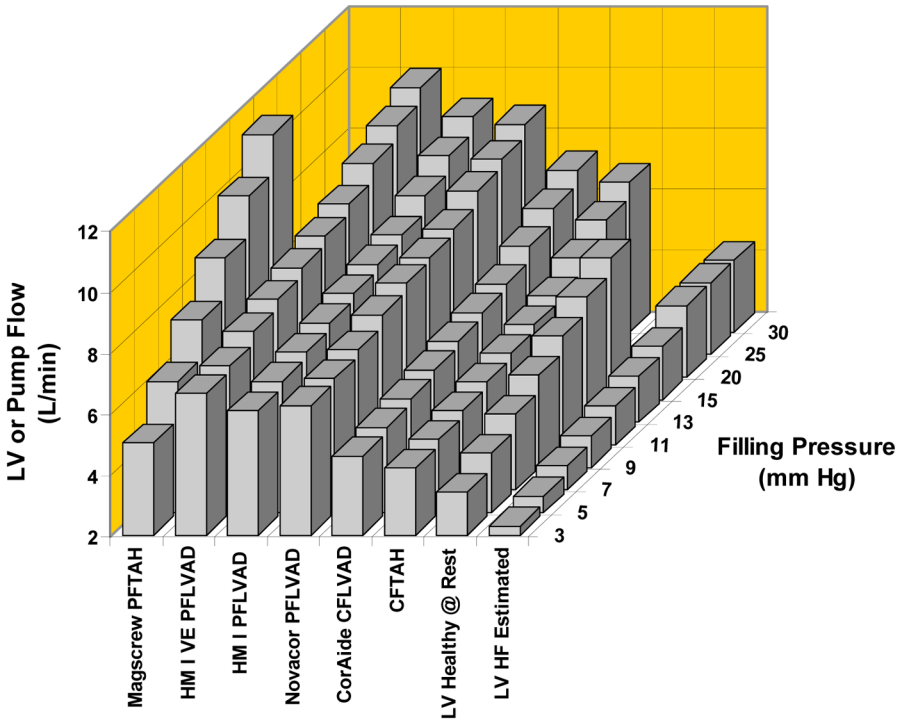

Figure 4.
Preload sensitivity relationships for the devices evaluated in our mock circulatory loops and that obtained for the human left ventricle.

Table 1

Preload Sensitivity of Mechanical Cardiac Assist Devices and the Human LV

Flow type	Pump type	Device Preload Sensitivity (L/min/mm Hg)		Device Class Preload Sensitivity (L/min/mm Hg)		
				Mean ± STD		
Continuous-flow devices	**CFLVAD**	CorAide	0.100	0.077 ± 0.040		0.080 ± 0.038
		HeartMate II [1, 7]	0.095			
		HeartWare HVAD [1, 6]	0.075			
		Jarvik 2000 [1, 7]	0.070			
		MicroMed DeBakey [8]	0.150			
		DuraHeart Terumo [1]	0.070			
		Incor Berlin Heart [1]	0.035			
		Abiomed Impella 2.5 [9]	0.020			
	CFTAH	CFTAH (Fixed speed)	0.102	0.102		
Pulsatile-flow devices	**PFLVAD** [AS]	HeartMate I Pneumatic	0.110	0.107 ± 0.015		0.537 ± 0.507
		HeartMate I XVE	0.120			
		Novacor	0.090			
	PFTAH [AS]	Utah-100 (Jarvik) AH [18]	1.400	0.900 ± 0.707	0.860 ± 0..436	
	PFTAH [FF]	Abiocor TAH [19]	0.400			
		MagScrew TAH [16]	0.640	0.820 ± 0.255		
		Penn State/3M TAH [15]	1.000			
Native ventricle	**Left Ventricle**	Human LV data (@ rest & 4 to 8 L/min) [21]	0.270	0.241 ± 0.040		
		Dog LV data (extrapolated to Starling curves) [1]	0.213			

AS = active suction; CF = continuous-flow; FF = free filling; LV = left ventricle; LVAD = left ventricular assist device; PF = pulsatile-flow; TAH = total artificial heart;.