Hemodynamic Support:
Science and Evaluation of the Assisted Circulation
with Percutaneous Assist Devices

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Hemodynamic Support:
Science and Evaluation of the Assisted Circulation with Percutaneous Assist Devices

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SYNOPSIS

Pathophysiological mechanisms that lead to hemodynamic abnormalities in cardiogenic shock (including hypotension, hypoperfusion and elevated venous pressures) are reviewed within the framework of pressure-volume analysis. This approach provides the foundation for understanding how different modes of circulatory support impact key cardiovascular parameters in various clinical settings. Four fundamentally different modes of circulatory support have been reviewed, including aortic counterpulsation, left atrial-to-arterial pumping, right atrial-to-arterial pumping and left ventricular-to-aortic pumping. Each approach has a distinct hemodynamic fingerprint with regard to effects on the ventricular pressure-volume loop and key hemodynamic parameters. Such understanding may help guide the choice of which device is most appropriate for a given clinical setting and also has the potential to guide future researchers in optimization of therapies, help generate hypotheses for clinical trials and help guide the choice of device for specific clinical trials that will ultimately provide evidence required to guide therapeutic decision making.
Introduction

The use of percutaneous devices to support the circulation in patients with various forms of hemodynamic compromise and for prophylactic use during high risk coronary interventions where such compromise is believed likely to occur is becoming more common. This is especially the case as such devices become easier to deploy, safer to use and hemodynamically more potent. As reviewed in the other chapters of this edition, the number of devices and the range of clinical indications in which they are applied are growing. In addition to devices to assist the left ventricle, the value of percutaneous right ventricular assist devices is becoming increasingly appreciated.

Since different devices have different modes of action, the clinician is faced with the task of choosing the appropriate device for each particular clinical setting. This can be facilitated by an understanding of fundamental hemodynamic principles that allow description of the nature and severity of changes in heart and vascular function in different disease states and the nature and potency with which different devices interact with the heart and vasculature. This is because the goals of cardiac support are different in different settings. For the elective setting of high risk coronary intervention, the goal is primarily to maintain reasonably normal systemic blood flow and blood pressure during a transient period of coronary occlusion and myocardial dysfunction. In contrast, in more emergent settings such as cardiogenic shock, pulmonary edema and pulmonary dysfunction, the goal is often to take over the work, partly or wholly, of the failing right and/or left ventricle to ensure normal blood pressure, cardiac output and pulmonary venous pressures over extended periods of time (1-5). When these goals are attained, end-organ perfusion and function are maintained, blood can be adequately oxygenated by
the lungs and diuresis is promoted in states of volume overload. Furthermore, by resting the heart and simultaneously ensuring end-organ perfusion in these settings, the odds of native heart recovery without permanent end-organ damage may be improved.

In addition to direct hemodynamic effects, the impact of different devices on coronary blood flow and myocardial oxygen demand can be important, especially in settings of acute coronary syndromes where preservation of myocardial function and viability are of primary concern in order to maximize the chances of recovery.

**Fundamental Hemodynamic Principles**

The ventricular pressure-volume framework provides a foundation for understanding cardiac and vascular properties, myocardial energetics and the impact of different modes of percutaneous circulatory support strategies. This framework allows representation of ventricular preload, afterload, lusitropy and contractility and their respective roles in determining cardiac output, blood pressure and pulmonary arterial pressures. Details of this approach have been summarized previously (6, 7) and will be reviewed here in brief. The basic concepts are summarized by the pressure-volume relations displayed in Fig. 1. The normal pressure-volume (PV) loop (shown in blue) is a plot of instantaneous ventricular pressure and volume throughout the cardiac cycle. The 4 major phases of the cardiac cycle are readily identified. Starting at end-diastole (bottom right corner) these phases are: isovolumic contraction, ejection, isovolumic relaxation and filling. The PV loop is bounded inferiorly by the end-diastolic pressure-volume relationship (EDPVR) and superiorly by the end-systolic pressure-volume

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*This and all other figures and quantitative values presented in this paper have been derived from a previously described and validated real-time, interactive cardiovascular simulation (8).*
relationship (ESPVR). The EDPVR uniquely defines the passive diastolic properties of the LV and the slope ($E_e$) and volume axis intercept ($V_o$) of the ESPVR provide a load-independent index of ventricular contractility.

In this construct, ventricular preload is indexed by either end-diastolic volume (EDV) or end-diastolic pressure (EDP). Ventricular afterload is indexed by effective arterial elastance ($E_a$) which is the slope of the line connecting the point on the volume axis at the end-diastolic volume to the end-systolic pressure-volume. $E_a$ is mainly determined by total peripheral arterial resistance (TPR) and the duration of the cardiac cycle ($T$) according to $E_a \equiv TPR/T$ (9).

One additional hemodynamic parameter of interest is cardiac power output (CPO) which is defined as the product of stroke work (SW, the area inside the PV loop) and heart rate (HR). However, since it is not possible to precisely quantify SW noninvasively, SW is approximated as the product of mean arterial pressure (MAP) and stroke volume (SV). Accordingly, CPO $\equiv$ SW $\cdot$ HR $\approx$ MAP $\cdot$ SV $\cdot$ HR $=$ MAP $\cdot$ CO. This parameter integrates information related to the two fundamental functions of the heart: its ability to generate blood pressure and to generate cardiac output. Additionally, interest in this parameter has stemmed largely from the fact that for patients presenting with cardiogenic shock, CPO has an inverse relation to 30 day survival (10-12).

**Myocardial Oxygen Demand and Supply**

Many clinicians are under the impression that myocardial oxygen consumption is related to SW (on a per beat basis) or to CPO (on a per unit time basis). However, this is not the case because SW does not quantify all of the work done by the heart with each
contraction. Myocardial oxygen consumption has been shown to be related to a parameter called the pressure-volume area (PVA) which is the sum of stroke work (the area inside the PV loop) and what is called the end-systolic potential area (PE) (13). PE is the area contained within the boundary defined by the ESPVR, the EDPVR and the diastolic portion of the PV loop (Fig. 2A) and represents residual energy stored in the myofilaments at the end of systole that is liberated as heat during crossbridge uncoupling. PVA is the total mechanical work done by the heart on a beat and is therefore closely related to myocardial oxygen consumption: \( PVA = SW + PE \) (14-16). The five pressure-volume loops shown in Fig. 2B are derived with a constant ventricular contractile state but with different values of total peripheral resistance. Fig. 2C shows the relationship between myocardial oxygen consumption (MVO2) for each of the loops as a function of the respective PVA values. When ventricular contractility is increased the MVO2-PVA relationship shifts upward in a parallel manner; conversely, when contractility decreases, the MVO2-PVA relationship shifts downward in a parallel manner (14). Since PVA relates to oxygen consumption \textit{per beat}, it should be appreciated that heart rate is a potent modulator of oxygen consumption per unit of time, since oxygen consumption per minute will be related to \( PVA \cdot HR \).

Oxygen is provided to the myocardium by blood delivered through the coronary arteries and microcirculation. Under normal conditions, the heart extracts more oxygen from blood than any other organ in the body, capable of attaining arterial-venous oxygen content differences of more than 15 ml O\(_2\)/100 ml blood, with typical arterial oxygen content of 20 ml O\(_2\)/100 ml blood. In addition, unlike other organs, the heart relies nearly completely on aerobic metabolism, necessitating oxygen to sustain myocardial
contraction. Therefore, under conditions of increased myocardial oxygen demand coronary there must be increased blood flow in order to maintain increased levels of total mechanical work. Coronary blood flow is regulated by metabolic, neural, humoral, autoregulatory, extravascular compressive and diastolic phase-related factors (17, 18). If coronary blood flow cannot increase to meet myocardial needs, increased workloads cannot be sustained. Similarly, when blood flow is limited (e.g., by pathological coronary occlusion by a thrombus or by occlusion during percutaneous intervention) myocardial workload decreases in an attempt to balance oxygen supply and demand. In general, the mechanism of reducing workload is accomplished by auto-downregulation of myocardial contractility in the affected (ischemic) region. If such a balance cannot be attained, myocardial necrosis ensues within short periods of time (e.g. on the order of 10 minutes).

If heart rate and epicardial resistance are fixed, as occurs with severe (though not total) coronary occlusion, blood flow can be increased mainly by either augmenting MAP, or by decreasing right atrial or LV end-diastolic pressure. Thus, in some settings, it is important to understand how circulatory support strategies impact MAP.

Cardiogenic Shock

In the discussions below, the impact of various percutaneous circulatory support strategies will be illustrated and compared in a state of acute cardiogenic shock (CGS) with pulmonary edema for which representative pressure-volume loops and relations are shown by the red lines in Fig. 1. With acute CGS as would occur in the setting of a large myocardial infarction, ventricular contractility is markedly reduced, as represented by the
downward shift of the ESPVR. In the acute setting, ventricular diastolic properties are not influenced significantly, so the EDPVR would be unchanged. The immediate effects of reduced contractility are decreased blood pressure (manifest as decreased height of the PV loop) and decreased stroke volume (decreased width of the PV loop). With activation of baroreflexes there are increases in heart rate (with a decrease in the duration of the cardiac cycle, T), increase in TPR and veno-constriction leading to a marked increase in left ventricular EDP and EDV. The increases in HR (i.e., decrease in T) and decrease in TPR result in a significant increase in Ea. All of these changes are readily identified in Fig. 1 by the PV loop and ESPVR in red. The resulting hemodynamic parameters for this example derived from the simulation in comparison to the normal condition are summarized in Table 1. Consistent with a state of CGS, systolic arterial pressure, cardiac output (CO) and thus CPO and PVA are all decreased and pulmonary capillary pressure is markedly increased.

Hemodynamic Effects of Different Percutaneous Support Strategies

The pressure-volume framework reviewed above is particularly useful for demonstrating and comparing the hemodynamic effects and metabolic consequences of different percutaneous support devices. The hemodynamic and metabolic impact of a support device depends on the flow rate of the pump and whether blood is pumped from the LV, LA or from the RA. The effects of pumping can also depend on the hemodynamic state from which support is initiated, which can vary from near normal (in the case of prophylactic use) to a state of deep cardiogenic shock. Although the first line therapy for all forms of hemodynamic compromise usually involves medical management
with inotropic agents and/or pressors (19, 20), these will not be discussed here because the focus of this paper is on hemodynamic effects of circulatory assist devices, not on the treatment of cardiogenic shock \textit{per se}. Nevertheless, it is noteworthy that while shown to increase blood pressure and cardiac output, the impact of such medical therapies on end organ perfusion can be variable (depending on the degree of peripheral vasoconstriction) and they have well-established adverse effects on the heart itself (1-5, 21, 22). Indeed, the combination of multiple agents appears to be associated with worse outcome (22). The effects of intravenous inotropes within the pressure-volume framework has been discussed previously (7).

This review will focus on the currently available forms of percutaneous circulatory support, including counterpulsation, extracorporeal circulatory support and intracorporeal trans-aortic valvular circulatory support.

\textit{Counterpulsation}

Counterpulsation with intra-aortic balloon pumping is often used in patients with otherwise untreatable myocardial ischemia (e.g., unstable angina) and also in patients with hemodynamic compromise as an adjunct to medical therapy (pressors and/or inotropes). A balloon with inflation volume of up to ~40cc alternately inflates during diastole and deflates during systole (Fig. 3A). Systolic balloon deflation is intended to reduce the pressure (and therefore the effective afterload) against which the heart ejects to improve cardiac output. Diastolic balloon inflation is intended to increase aortic pressure to increase coronary blood flow and end-organ perfusion. The theoretical effects of counterpulsation on the PV loop and hemodynamics are summarized in Fig. 3B and Table 1. As seen, the effects on parameters like cardiac output (~10%) and
pulmonary capillary pressure (~1 mmHg) are small (and may be beyond the sensitivity of detection in the clinical setting) while the effects of counterpulsation on coronary pressure, and therefore coronary flow, can be significant. Consistent with the conclusions, results of several clinical studies have confirmed no significant hemodynamic effectiveness of counterpulsation in cardiogenic shock (23-26). From an energetic standpoint, there is also a small (~3%) reduction in PVA implying that counterpulsation would not significantly reduce myocardial oxygen demands. Yet, there can be a significant increase in CPO owing to the increase in blood pressure. Nevertheless, the results of a recent large randomized study showed no survival benefit from counterpulsation in cardiogenic shock (27).

Extracorporeal Left Atrial-to-Arterial Circulatory Support

Percutaneous LA-to-arterial circulatory support devices (e.g., as with the TandemHeart device) have significant beneficial effects on hemodynamic parameters, as illustrated in Fig. 4 and Table 1 and shown in prior clinical studies (28, 29). Such devices have flow capacities approaching 4 L/min and since they draw blood directly from the left atrium, pulmonary capillary pressure is reduced significantly, by 4 mmHg in the present example. Accordingly, LV end-diastolic volume and pressure are also decreased because blood is diverted from flowing through the mitral valve. At the same time, there is now a continuous flow of blood to the aorta throughout the cardiac cycle, even during diastole. This has the effect of increasing aortic diastolic pressure and decreasing aortic pulse pressure (PP, which is systolic minus diastolic pressure; Fig. 4A). As a consequence of the decrease in LV preload and increase in LV afterload, intrinsic cardiac
output from the LV is decreased substantially. Thus, although cardiac output was 3.5 L/min prior to initiation of support and the extracorporeal system is pumping 3.3 L/min, the final total output during support is not 3.5+3.3=6.8 L/min, but is 4.4 L/min, an increase of only 1.1 L/min. This is because intrinsic output from the LV decreased from 3.5 to 1.1 L/min due to the effects of the assist device. The reduction of cardiac output is reflected as a decrease in stroke volume (the width of the PV loop) and, accordingly, a reduction in ejection fraction (EF). This example therefore illustrates two important and fundamental principles of circulatory support:

1. The final total cardiac output achieved after implantation of partial support circulatory assist device is not the sum of the original cardiac output plus the flow of the assist device. The final cardiac output depends on many factors and, as discussed previously, no general statement can be made as to the expected increase in CO following initiation of circulatory support (30). In addition, it should specifically be noted that the total cardiac output cannot be determined through examination of the PV loop. Naturally, when the assist device is powerful enough to overtake and pump more than the native heart, the assist device alone determines the total flow.

2. When such a device is used, the marked changes in pre- and afterload that result in reductions in intrinsic stroke volume and cardiac output can result in a decrease in EF. Such a reduction does not reflect a reduction in ventricular contractility. As shown in Fig. 4 the end-systolic pressure-volume point falls on the same ESPVR indicating the
constancy of LV contractility. The change in EF is purely a result of the change in loading conditions induced during circulatory assist.

In addition to the impact on pulmonary capillary pressure, cardiac output and diastolic blood pressure, the increase in total cardiac output also results in an overall increase in blood pressure throughout the cardiac cycle which has the potential benefits of improving coronary blood flow and end-organ perfusion. This approach also results in a substantial increase in CPO, frequently to almost normal values. On the other hand, there is little impact on PVA (in particular no significant reduction), principally because of the significant increase in arterial pressure. Accordingly, it is expected that there is little effect on myocardial oxygen consumption with this strategy.

*Extracorporeal Right Atrial-to-Arterial Circulatory Support*

Right atrial-to-arterial circulatory assist devices are becoming increasingly used for patients with severe hemodynamic compromise as with biventricular failure and with pulmonary dysfunction, typically in combination with membrane oxygenation and sometimes with a heat exchanger. Specialized catheters for venous and arterial access are readily available for this application. Regardless of configuration, these systems are generically referred to as extracorporeal membrane oxygenation (ECMO) systems. The pumps typically used in these circuits are powerful and capable of overtaking the native heart, but are generally set at ~4 L/min. There are distinct hemodynamic differences between sourcing the blood from the right atrium as opposed to the left atrium which was discussed in the prior section. The major difference, readily appreciated from the PV

* Venous-to-venous pumping configurations, used in patients in whom oxygenation or hypercapnea is the primary problem instead of hemodynamic compromise, will not be discussed in this paper.
loop (Fig. 5 and Table 1), is that when applied to patients with compromised LV function this configuration can result in further significant and potentially detrimental increases in pulmonary capillary and left ventricular end-diastolic pressures and LV distension can results. Because of this, some form of LV venting may be required, though not generally able to be implemented percutaneously (see further discussion of this below). This LV loading effects is due to the fact that with RA-to-arterial circulatory support the only path for blood to leave the LV is via the aortic valve and in order to overcome the significantly elevated arterial pressure in the setting of the decreased contractility, that can only occur at high (higher than starting) levels of ventricular preload. Accordingly, whereas with LA-to-arterial pumping the higher the flow rate the greater the degree of ventricular unloading, with RA-to-arterial pumping, the higher the flow rate the greater the degree of ventricular loading. These increases in preload result in further increases PVA and, therefore, oxygen consumption. Confirmation of this loading effect of ECMO was recently reported by Kawashima who noted consistent elevation of PVA in an animal model of varying degrees of LV failure (26).

*Intracorporeal Trans-aortic valve Circulatory Support*

The final class of percutaneous circulatory support devices to be considered are pumps placed at the ends of catheters that source blood from the LV and pump it to the aorta. Although the Hemopump was the first device developed specifically with this mode of action, technical problems prevented that particular device from being a viable clinical product (31). More recently, the Impella class of devices has been introduced and are now being used on a routine basis (32). Three different sized pumps are available,
each with its own maximum pumping capacity: 2.5, 4.0 and 5.0 L/min. Since all three devices employ an LV-to-aorta support strategy, the impact on hemodynamics and myocardial energetics are fundamentally the same, becoming more “potent” in both respects as the flow rate is increased.

The impact of LV-to-aorta pumping on the PV loop, illustrated in Fig. 6B (with pumping rates of 2.4 L/min in green, 3.5 L/min in orange and 4.75 L/min in magenta), is fundamentally different than the other support strategies. Specifically note that because these devices are pumping blood continuously out of the LV into the aorta independent of the phase of the cardiac cycle, there are no isovolumic contraction or isovolumic relaxation periods; the loop transforms from a more or less rectangular shape to a triangular shape, especially at the highest pumping rates. Note also that as pumping speed is increased, there is a progressively greater leftward shift of the loop to lower ventricular EDVs and EDPs, corresponding with progressively smaller PVAs and, therefore, progressively lower levels of myocardial oxygen demand.

As with the other support systems, the impact on CO cannot be determined from the PV loop since both heart and device contribute to the total flow to the body. As summarized in Table 1, native CO flow decreases as pump flow increases but total flow and MAP increase, which has the potential to improve coronary (33) and end-organ perfusion. With the device pumping 2.4 L/min, native CO decreases by ~1/3 but total output increases by ~0.5 L/min and there is an ~10 mmHg increase in MAP. As shown in Fig. 6A, the increase in MAP is due almost entirely to an increase in diastolic pressure with very little increase in systolic pressure; accordingly, PP decreases. The rise in
diastolic pressure is due to continual pumping of blood from the LV to aorta during diastole.

With pump flow increased to 3.5 L/min and the further reduction in EDV and EDP, total flow is increased by 0.7 L/min above the starting cardiogenic shock value as native CO decreases to only 0.65 L/min. MAP increases further, again due to further increased diastolic pressure with a relatively small increase in systolic pressure and a further reduction in PP.

Finally, with the device pumping 4.75 L/min, the native heart is overcome and all of the flow is from the pump. The LV cannot generate pressure to overcome arterial pressure the aortic valve stays closed. Accordingly, the arterial pressure losses pulsatility, with only minimal fluctuations from slight variations in pump output due to time-dependent changes in LV-to-aorta pressure gradients during the cardiac cycle. It is appreciated on the PV diagram that despite the fact that the heart is not ejecting into the aorta, there are still cycle dependent changes in ventricular volume owing to pressure-dependent opening and closing of the mitral valve in response to ventricular contraction and pressure variations.

Other Considerations

The discussions above have been based around results obtained from a cardiovascular simulation (8, 30). Although validated to a certain extent, the behavior of the real cardiovascular system is significantly more complex than can be captured in any such simulation. Furthermore, there is significant variability in hemodynamic characteristics from patient-to-patient. Even then, the predicted effects do not account for
changes in patient status following initiation of circulatory support that can be mediated by baroreflexes, changes in renal function, the multitude of drugs that are used in different clinical scenarios and many other factors. The presence of valvular lesions, not considered at all, can have a profound effect on device performance. Finally, only one particular hemodynamic profile of cardiogenic shock has been considered; there are infinite possible combinations of ventricular and vascular dysfunction with which patients present. As emphasized at the start, the impact of any circulatory support system is highly dependent on native cardiovascular properties. As just one example, the impact of circulatory support in cardiogenic shock in the absence of profound elevations in pulmonary capillary pressure is quantitatively very different with regard to changes in ventricular EDV and EDP achieved with the different support strategies.

The hemodynamic demands surrounding the prophylactic use temporary percutaneous support are significantly more modest than those related to their use in cardiogenic shock, both in terms of hemodynamic and duration of use requirements. These have been discussed previously (34) and have not been reviewed here.

Finally, although the fundamental principles underlying the use of such devices for right ventricular support are similar, there are many unique features of such applications, especially as they relate to the nature of the underlying hemodynamic abnormalities being treated. These factors, in combination with several important differences between right and left ventricular physiology caution against simple extrapolation of concepts discussed above to apply to right-sided support.

Taking all of this into account, one should consider that the information reviewed above provides a framework for understanding basic hemodynamic concepts of
circulatory support and not that it provides generally applicable quantitative information about the hemodynamic effects of any particular circulatory support device in any particular patient. In order to take this approach to that level, one potentially fruitful line of investigation is to further develop such computer simulations to be able to model specific patients and be able to predict the hemodynamic effects of specific interventions: i.e., development of algorithms for pathophysiology-based personalized medicine. Initial work on such an approach has already been laid down (30) but requires significant further development and validation.

**Summary**

Four different approaches to circulatory support have been reviewed. Each approach has a distinct hemodynamic fingerprint with regard to effects on the ventricular pressure-volume loop, cardiac output, arterial pressure, pulmonary capillary pressure, pressure-volume area and myocardial oxygen consumption. An understanding of basic hemodynamic concepts and mechanisms the lead to hemodynamic compromise (including hypotension, hypoperfusion and elevated venous pressures) provides a foundation for understanding how different modes of circulatory support impact key cardiovascular parameters in various clinical settings. Such understanding can help guide the choice of which device is most appropriate for a given patient.

This is particularly important because in the field of percutaneous circulatory support there are very few randomized clinical trials to guide therapeutic decisions. With the lack of such evidence to establish treatment guidelines, therapeutic decision making is based on first principles and, ultimately, experience.
In this regard, there are several key take away messages. First, consistent with prior clinical trials (29, 35) the hemodynamic effects of counterpulsation appear to be mainly restricted to effects on blood pressure. Whether this results in any clinical secondary benefits beyond reducing myocardial ischemia is unknown (27).

Most active pump circulatory support systems provide partial support in that they do not completely take over heart function. Rather, these devices generally work in concert with the native heart. When operating in this mode, the final total CO is not the sum of the cardiac output prior to initiation of support plus the flow of the device. The resulting changes in LV preload and afterload generally reduce intrinsic CO and the final total CO is determined by the complex interactions between heart, vasculature and device.

When the device overtakes the intrinsic pumping capacity of the heart, the aortic valve remains closed and CO is determined exclusively by the heart.

With an LA-to-arterial support strategy, the greater the amount of pumping the greater the increase in MAP and the greater the reduction in ventricular preload. With an RA-to-arterial strategy, the greater the amount of pumping the greater the increase in MAP but the greater the potential for increasing ventricular preload. With an LV-to-aorta strategy, the more you pump the more you reduce both ventricular preload and MAP; minimization of increases in systolic arterial pressure with this approach in comparison to the LA-to-arterial pumping strategy appears to make this approach more effective in reducing PVA and, potentially, myocardial oxygen consumption.

Although some of these conclusions may seem trivial, it is an understanding of the underlying theories that have the potential to guide future researchers in optimization of therapies, help generate hypotheses for clinical trials and help guide choice of the
proper device for specific clinical trials that will ultimately provide evidence required to guide therapeutic decision making.
Reference List


FIGURE LEGENDS

**Figure 1:** Prototypical pressure-volume (PV) loops from a normal adult (blue) and from a patient in cardiogenic shock (CGS, red). The loops are bound by the end-systolic and end-diastolic pressure-volume relations (ESPVR and EDPVR, respectively). The downward shift of the ESPVR is a reflection of the reduction in left ventricular contractility that underlies the development of CGS. Ea, an index of ventricular afterload which depends primarily on total peripheral resistance and heart rate, can also be derived from the loop. See text for further details.

**Figure 2:** A. Myocardial oxygen consumption is proportional to pressure-volume area (PVA) which is the sum of external stroke work (SW) and end-systolic potential energy (PE). At a given contractile state, when PVA is varied by changes in loading condition (as in panel B), there is a linear relationship between PVA and myocardial oxygen consumption (MVO2) on a per beat basis.

**Figure 3:** Theoretical impact of counterpulsation with a 40 cc intra-aortic balloon pump on arterial pressure (A) and pressure volume loop (B) in cardiogenic shock (CGS). Baseline CGS hemodynamic state described in Fig. 1 (shown in red) are used as starting point to assess the effects of different circulatory support strategies. During counterpulsation (shown in green) balloon deflation decreases blood pressure during ejection (reduced effective ventricular afterload) and balloon inflation during diastole increases arterial pressure with the goal of improving coronary and end-organ perfusion.
The impact of counterperfusion on the pressure-volume loop (panel B) shows the reduction in peak ventricular pressure and relatively small effect on stroke volume and end-diastolic volume. Refer to Table 1 for more detailed summary of hemodynamic effects.

**Figure 4:** Impact of left atrial-to-arterial circulatory support strategy on arterial pressure (A) and pressure-volume loop (B). This approach creates a significant increase in diastolic more than systolic arterial pressure (with reduction in pulse pressure). There is a significant reduction in LV end-diastolic volume and pressure. Pressure-volume area, however, is not significantly altered because of the offsetting effects of the increase in afterload and reduction in preload. Refer to Table 1 for more detailed summary of hemodynamic effects.

**Figure 5:** Impact of right atrial-to-arterial circulatory support strategy on arterial pressure (A) and pressure-volume loop (B). This approach creates a significant increase in diastolic more than systolic arterial pressure (with reduction in pulse pressure). There is a significant increase in LV end-diastolic volume and pressure. Consequently, there is ventricular overloading and pressure-volume area increases. Refer to Table 1 for more detailed summary of hemodynamic effects.

**Figure 6:** Impact of left ventricular-to-arterial circulatory support strategy on arterial pressure (A) and pressure-volume loop (B). Three levels of support are shown: 2.4 L/min (green), 3.5 L/min (orange) and 4.75 L/min (magenta). This approach creates a
significant increase in diastolic pressure with little impact on systolic arterial pressure until the degree of support is sufficient to overtake the natural heart. There is a significant reduction in LV end-diastolic volume and pressure that increases as the magnitude of support increases. Note that the pressure-volume loop loses its normal rectangular shape in favor of an increasingly triangular shape as the magnitude of support increases. Pressure-volume area decreases in relation to the amount of flow from the device. Refer to Table 1 for more detailed summary of hemodynamic effects.
Table 1. Hemodynamic parameters derived from a cardiovascular model simulating normal conditions, cardiogenic shock and then cardiogenic shock with addition of different types of circulatory support devices.

<table>
<thead>
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<th>Hemodynamic Parameter</th>
<th>Normal</th>
<th>CGS</th>
<th>+IABP</th>
<th>LA → Ao 3.3 L/min</th>
<th>LV → Ao 2.4 L/min</th>
<th>LV → Ao 3.5 L/min</th>
<th>LV → Ao 4.75 L/min</th>
<th>RA → Ao 4.0 L/min</th>
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<tr>
<td>PA Mean</td>
<td>16</td>
<td>32</td>
<td>31</td>
<td>29</td>
<td>31</td>
<td>30</td>
<td>28</td>
<td>37</td>
</tr>
<tr>
<td>PCP</td>
<td>13</td>
<td>30</td>
<td>29</td>
<td>26</td>
<td>28</td>
<td>27</td>
<td>25</td>
<td>36</td>
</tr>
<tr>
<td>Ao Systolic</td>
<td>110</td>
<td>92</td>
<td>88</td>
<td>102</td>
<td>95</td>
<td>96</td>
<td>102</td>
<td>110</td>
</tr>
<tr>
<td>Ao Diastolic</td>
<td>58</td>
<td>67</td>
<td>97</td>
<td>91</td>
<td>81</td>
<td>88</td>
<td>101</td>
<td>99</td>
</tr>
<tr>
<td>Ao Mean</td>
<td>78</td>
<td>77</td>
<td>86</td>
<td>95</td>
<td>86</td>
<td>90</td>
<td>101</td>
<td>103</td>
</tr>
<tr>
<td>CPO (Watts)</td>
<td>0.92</td>
<td>0.60</td>
<td>0.74</td>
<td>0.94</td>
<td>0.76</td>
<td>0.83</td>
<td>1.07</td>
<td>1.14</td>
</tr>
<tr>
<td>PVA (mmHg.ml)</td>
<td>10,363</td>
<td>7,850</td>
<td>7,637</td>
<td>7,878</td>
<td>7,537</td>
<td>7,290</td>
<td>6,860</td>
<td>8,700</td>
</tr>
</tbody>
</table>

CGS, cardiogenic shock; IABP, intraaortic balloon pump; LA, left atrium; LV, left ventricle; Ao, aorta; RA, right atrium; PA, pulmonary artery; PCP, pulmonary capillary pressure; CPO, cardiac power output; PVA, pressure-volume area.
FIGURE 2

A

\[
\begin{align*}
\text{Pressure (mmHg)} & \quad \text{Volume (ml)} \\
0 & \quad 0 \\
125 & \quad 120 \quad \text{SW} \\
25 & \quad 40 \\
0 & \quad 40 \\
0 & \quad 120
\end{align*}
\]

B

\[
\begin{align*}
\text{Pressure (mmHg)} & \quad \text{Volume (ml)} \\
0 & \quad 0 \\
180 & \quad 130 \quad \text{ESPVVR} \\
120 & \quad 90 \\
0 & \quad 52 \\
120 & \quad 78 \\
0 & \quad 104
\end{align*}
\]

C

\[
\begin{align*}
\text{MVO2 (mL O2/beat)} & \quad \text{Pressure-Volume Area (mmHg.ml)} \\
0.05 & \quad 0 \\
0.10 & \quad 3000 \\
0.15 & \quad 6000 \\
0.10 & \quad 9000 \\
0.05 & \quad 12000
\end{align*}
\]

\[
\text{SW + PE = PVA}
\]
FIGURE 3
FIGURE 5

A

AoP (mmHg)

110

100

90

80

70

60

0.5 sec

B

Pressure (mmHg)

100

75

50

25

Volume (ml)

140

160

180