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Computational Evaluation of IABP, Impella 2.5, TandemHeart and Combined IABP and Impella 2.5 Support in Cardiogenic Shock

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Abstract: Cardiogenic shock is a life-threatening condition consisting of low cardiac output status leading to end-organ hypoperfusion following either acute left or right ventricular failure or decompensation of chronic heart failure. Partial or failed response to inotropic support in the acute phase may require the use of mechanical circulatory support. Although patients supported with different devices such as an IABP, Impella 2.5, or TandemHeart experience stability in the short term, the haemodynamic benefits of each device remain unclear. The aim of this study is to present a direct comparison of an IABP, Impella 2.5, TandemHeart, and combined IABP and Impella 2.5 support in cardiogenic shock to evaluate haemodynamic variables and left ventricular unloading using cardiovascular system modelling and simulation in terms of cardiac function, systemic, pulmonary, cardiac, and cerebral circulations. The simulation results showed that the IABP had a relatively low effect on the haemodynamic variables. Although both Impella 2.5 and TandemHeart improved the total blood flow rates, as well as coronary and cerebral perfusion with the increasing pump operating speed, TandemHeart had a more profound effect on the haemodynamic variables. Combining the IABP and Impella 2.5 also improved the haemodynamics, although at the expense of reverse blood flow in the cerebral circulation. Simulation results showed that TandemHeart support might have a more beneficial effect on the haemodynamics and left ventricular energetics in comparison to the IABP and Impella 2.5. Nevertheless, the combined use of the IABP and Impella 2.5 for short-term support may be considered an appropriate alternative.

Keywords: cardiovascular system model; IABP; Impella 2.5; TandemHeart; cardiogenic shock; haemodynamics

MSC: 37M05

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1. Introduction

Cardiogenic shock (CS) is a life-threatening condition consisting of low cardiac output status leading to end-organ hypoperfusion following either acute left or right ventricular failure or decompensation of chronic heart failure [1]. Initial medical management in terms of inotropic support, vasopressors, and diuretics aims to restore cardiac output and reduce the potential for irreversible myocardial damage [2]. Partial or failed response to inotropic support in the acute phase may require the use of mechanical circulatory support (MCS) to maintain end-organ function in these patients [3]. The aim of MCS in CS is to increase systemic perfusion, enhance coronary blood flow, reduce left ventricular filling pressures, and decrease myocardial oxygen consumption [4]. MCS devices such as the Intra-Aortic Balloon Pump (IABP), Impella 2.5, and TandemHeart play a crucial role in managing CS and supporting the failing heart in the short-term [5].

The IABP is composed of an expandable balloon that is inserted percutaneously via the femoral artery and positioned slightly below the point where the left subclavian artery originates [6]. This placement is achieved either by using fluoroscopic guidance within the catheterisation laboratory or with the assistance of transoesophageal echocardiography guidance in the operating theatre [6]. Synchronisation with the cardiac cycle allows inflation during diastole and deflation during systole [7]. The counterpulsation mechanism of the IABP aims to improve coronary perfusion and reduce the cardiac workload [8]. It provides temporary circulatory support during and after surgical procedures, particularly in patients at high risk for postoperative complications [9].

Impella 2.5 is a percutaneous left ventricular assist device that provides temporary MCS [10]. It is a 12-Fr catheter system inserted through the femoral artery and positioned across the aortic valve under fluoroscopic guidance in the cath lab, allowing blood to be pumped from the left ventricle into the aorta [10]. Impella 2.5 support can unload the left ventricle, improve coronary perfusion, and increase systemic blood flow [11].

TandemHeart is a continuous-flow centrifugal extracorporeal assist device operating in parallel with the left ventricle. Oxygenated blood is withdrawn from the left atrium and returned to the outflow cannula placed into the femoral artery. The inflow cannula is advanced through the femoral vein and placed trans-septally into the left atrium under fluoroscopic guidance in the cath lab [12]. The TandemHeart is used in severe CS refractory to conventional therapy [13].

The IABP is the most used device in CS patients [14], although 12-month mortality rates in CS patients are high [15]. Also, the benefits of Impella 2.5 support on the survival rates in CS patients remain unclear. For instance, Lauten et al. [16] reported that the overall mortality after Impella 2.5 insertion was around 64%. However, another study concluded that Impella 2.5 support improved survival rates in CS patients [17]. A comparative study showed that there was no difference in 30-day mortality rates in patients supported with Impella 2.5 or an IABP [18]. TandemHeart improves the cardiac index, systolic blood pressure, urine output, and lactic acid level in CS patients [19]. However, mortality rates in patients supported with TandemHeart remain similar to patients requiring IABP insertion [20]. Impella and Veno-Arterial Extra Corporeal Membrane Oxygenation (VA-ECMO) were also combined to improve haemodynamic support in CS [21]. Even though combined Impella and VA-ECMO may reduce mortality, it is still unclear whether all patients may benefit from this combined device approach [21].

Computational models and simulations can provide valuable insights into the haemodynamic outcome of MSC. However, most of the previous studies focused on the modelling of MSC devices rather than comparing the benefits of different MCS systems and configurations. For instance, Abdolrazaghi et al. [22] developed a mathematical model to simulate IABP support. Sun [23] modelled and simulated the interaction of the left ventricle and IABP. Schampaert et al. [24] evaluated the timing of IABP-triggering using in silico and in vitro models. De Lazzari et al. [25] evaluated the effect of IABP vacuum pressure on left ventricular unloading. Yu et al. [26] developed a model to simulate the hydraulic characteristics of TandemHeart. Although these studies provide information about the outcome of different MCS for different devices, a direct comparison of different MCS systems and configurations may help evaluate the unloading of the left ventricle for better support in CS. Therefore, the aim of this study is to present a direct comparison of an IABP, Impella 2.5, TandemHeart, and combined IABP and Impella 2.5 support in CS to evaluate haemodynamic variables and left ventricular unloading using numerical simulations.

2. Materials and Methods

The numerical model that was employed replicates the functioning of the heart as well as the systemic, pulmonary, and cerebral circulatory systems. The simulation takes into account the left ventricular pressure (p_{lv}), encompassing both active and passive elements ($p_{lv,a}$ and $p_{lv,p}$).

$$p_{lv} = p_{lv,a} + p_{lv,p}, \tag{1}$$

The contraction of the left ventricle was initiated by a function ($f_{act,lv}$), while the end-systolic elastance ($E_{es,lv}$) and the volume of the left ventricle (V_{lv}) were employed to derive the active pressure of the left ventricle ($p_{lv,a}$).

$$p_{lv,a}(t) = E_{es,lv}(V_{lv} - V_{lv,0})f_{act,lv}(t)\ , \tag{2}$$

The passive pressure of the left ventricle ($p_{lv,p}$) was simulated through the utilization of an exponential function that incorporates both the volume of the left ventricle (V_{lv}) and supplementary parameters (A_{lv}, B_{lv}).

$$p_{lv,p} = A_{lv}\left[e^{B_{lv}(V_{lv}-V_{lv,0})} - 1\right], \tag{3}$$

The volume of the left ventricle (V_{lv}) was regarded as a function of the left ventricular radius (r_{lv}), the length of the long axis (l_{lv}), and a scaling parameter (K_{lv}).

$$V_{lv} = \frac{(4/3)\pi K_{lv} r_{lv}^2 l_{lv}}{2}, \tag{4}$$

The rate of change of the left ventricular radius with respect to time (dr_{lv}/dt) was formulated using the left ventricular volume (V_{lv}), the length of the left ventricular long axis (l_{lv}), the scaling coefficient (K_{lv}), and the flow rates passing through the mitral and aortic valves (Q_{mv}, Q_{av}).

$$\frac{dr_{lv}}{dt} = \frac{3(Q_{mv} - Q_{av})}{4\pi K_{lv} l_{lv}}\left(\frac{6V_{lv}}{4\pi K_{lv} l_{lv}}\right)^{-1/2}, \tag{5}$$

The left atrial pressure was simulated by incorporating the left atrial elastance function ($E_{la}(t)$), left atrial volume, and zero pressure–volume (V_{la}, $V_{la,0}$).

$$p_{la}(t) = E_{la}(t)(V_{la} - V_{la,0}), \tag{6}$$

The volume of the left atrium (V_{la}) depended on the left atrial radius (r_{la}), its long axis length (l_{la}), and a scaling parameter (K_{la}). The change rate of the left atrial radius over time (dr_{la}/dt) was modelled using the flow rates from the pulmonary veins and mitral valve (Q_{vp}, Q_{mv}), the left atrial volume (V_{la}), its long axis length (l_{la}), and the coefficient, K_{la}.

$$V_{la} = \frac{2}{3}\pi K_{la} r_{la}^2 l_{la}, \tag{7}$$

$$\frac{dr_{la}}{dt} = \frac{3(Q_{vp} - Q_{mv})}{4\pi K_{la} l_{la}}\left(\frac{3V_{la}}{2\pi K_{la} l_{la}}\right)^{-1/2} \tag{8}$$

Detailed information about Equations (1)–(8) can be found in [27]. The functions of the right atrium and ventricle were replicated in a similar manner, but the parameter values within these compartments were different. The heart valves permitted unidirectional blood flow, with blood flow rates determined by the pressure differential across the valve and valve resistances (R). The flow rate through the mitral valve (Q_{mv}) is presented below.

$$Q_{mv} = \frac{p_{la} - p_{lv}}{R_{mv}}, \tag{9}$$

Blood circulation was elucidated through a 0D model, encompassing an electrical analogy to represent the resistance (R), compliance (C), and inertia (L) within the blood vessels. Alterations in aortic blood pressure and flow rate with respect to time (dp_{ao}/dt, dQ_{ao}/dt) are provided below.

$$\frac{dp_{ao}}{dt} = \frac{Q_{av} - Q_{ao}}{C_{ao}}, \tag{10}$$

$$\frac{dQ_{ao}}{dt} = \frac{p_{ao} - p_{as} - R_{ao}Q_{ao}}{L_{ao}}, \tag{11}$$

Here, Q_{av} and p_{as} denote the flow rate through the aortic valve and the systemic arteriolar pressure, respectively. In Equation (11), C_{ao}, R_{ao}, and L_{ao} stand for aortic compliance, resistance, and inertance.

The cerebral circulation includes the main cerebral arteries and the Circle of Willis. Detailed information about cardiac function and cerebral circulation modelling can be found in [27–29]. Cardiac circulation includes left and right coronary arteries, coronary arterioles, coronary capillaries, and the coronary sinus. Left and right coronary arterial resistances were simulated using piecewise equations, which set the left and right coronary arterial resistances to upper values over the ejection phase and lower values after the aortic valve closes.

The regulation of systemic arteriolar resistance (R_{ars}) was accomplished by employing the mean aortic pressure ($p_{ao,m}$) within the cardiovascular system model, as outlined in reference [30].

$$\Delta R_{ars} = |\, S_{Rars}(p_{ao,ars,set} - p_{ao,m}) R_{ars,set} | \tag{12}$$

$$R_{ars} = \begin{cases} R_{ars} - \Delta R_{ars} & p_{ao,m} \geq p_{ao,ars,set} \\ R_{ars} + \Delta R_{ars} & p_{ao,m} < p_{ao,ars,set} \end{cases} \tag{13}$$

Here, S_{Rars}, ΔR_{ars}, $R_{ars,set}$, and $p_{ao,ars,set}$ represent the sensitivity coefficient, change in the resistance, resistance at the set point, and aortic pressure set point, as described in [28,30].

CS was simulated by reducing the left ventricular systolic elastance ($E_{es,lv}$) from 2.5 mmHg/mL to 0.75 mmHg/mL and parameter A (A_{lv}) in Equation (3) from 1 to 0.65. Circulatory support was simulated by implementing numerical models of Impella 2.5, the IABP, and TandemHeart into the cardiovascular system network. Also, the haemodynamic outcome of combined Impella 2.5 and IABP support was simulated.

Impella 2.5 support was simulated using a numerical model [25], which described pressure and flow rate relations across the pump, as given below.

$$Q_{IMP} = K_1(p_{ao} - p_{lv})^4 + K_2(p_{ao} - p_{lv})^3 + K_3(p_{ao} - p_{lv})^2 + K_4(p_{ao} - p_{lv}) + K_5 \quad (14)$$

Here, Q_{IMP} represents the flow rate through an Impella 2.5 device, p_{ao} and p_{lv} are the aortic and left ventricular pressures, and K_i (i = 1–5) are constant coefficients changing at different pump operating speeds. Values of K_i at different pump operating speeds and detailed information about the Impella 2.5 model can be found in [25].

IABP support was simulated using varying resistance [24] and a pressure source, which represents varying IABP pressures at the insertion site of the balloon, as given below.

$$\frac{d(p_{aa} - p_{IABP})}{dt} = \frac{Q_{ao} - Q_{aa} - Q_{lica} - Q_{rica} - Q_{lva} - Q_{rva}}{C_{aa}} \quad (15)$$

$$R_{aa} = R_{base,aa}\left[1 - \left(\frac{r_{IABP}}{r_{aa}}\right)^4 - \frac{\left(1 - (r_{IABP}/r_{aa})^2\right)^2}{\ln(r_{IABP}/r_{aa})}\right]^{-1} \quad (16)$$

Here, r_{IABP} and r_{aa} represent the radius of the IABP and the insertion site. Detailed information about the IABP model can be found in [24]. IABP pressure was described assuming linear variations between vacuum and drive pressures over an 80 ms period, as described in [24]. Combined Impella 2.5 and IABP support was simulated by implementing the models of both devices.

TandemHeart support was simulated using a model that described static pressure and flow rate characteristics of the device at different operating speeds [26].

$$\Delta p_{TH} = P_{TH}(n) - R_{p0}(n)Q_{TH} - R_{p1}(n)|Q_{TH}|Q_{TH} \quad (17)$$

Here, Δp_{TH} and Q_{TH} represent the pressure difference across the TandemHeart and pump flow rate. $P_{TH}(n)$, R_{p0}, and R_{p1} are the variables depending on the pump operating speed. Detailed information about the TandemHeart model and pump parameters can be found in [26].

CS was simulated by tuning the left ventricular end-systolic elastance, as described in [27,28]. The IABP was operated by adjusting the vacuum pressure to 0 mmHg–10 mmHg and 10 mmHg and 260 mmHg drive pressures, as described in [25]. Impella 2.5 was simulated by operating the pump at 40,000 rpm, 45,000 rpm, and 50,000 rpm speeds using the pump parameters in [25]. TandemHeart support was simulated by operating the pump at 3000 rpm, 4500 rpm, and 6000 rpm speeds using the pump parameters in [26]. Combined IABP and Impella 2.5 support was simulated by operating the IABP at 0 mmHg vacuum and 206 mmHg drive pressures and Impella at a 40,000 rpm speed. The electric circuit representation of the cardiovascular system and MCS is given in Figure 1.

Matlab Simulink R2021b was used to run the simulations. The entire model was simulated using the ode15s solver. The maximum step size was 5×10^{-4} s, and the relative tolerance was 5×10^{-4}. The heart rate was 75 bpm in the simulations.

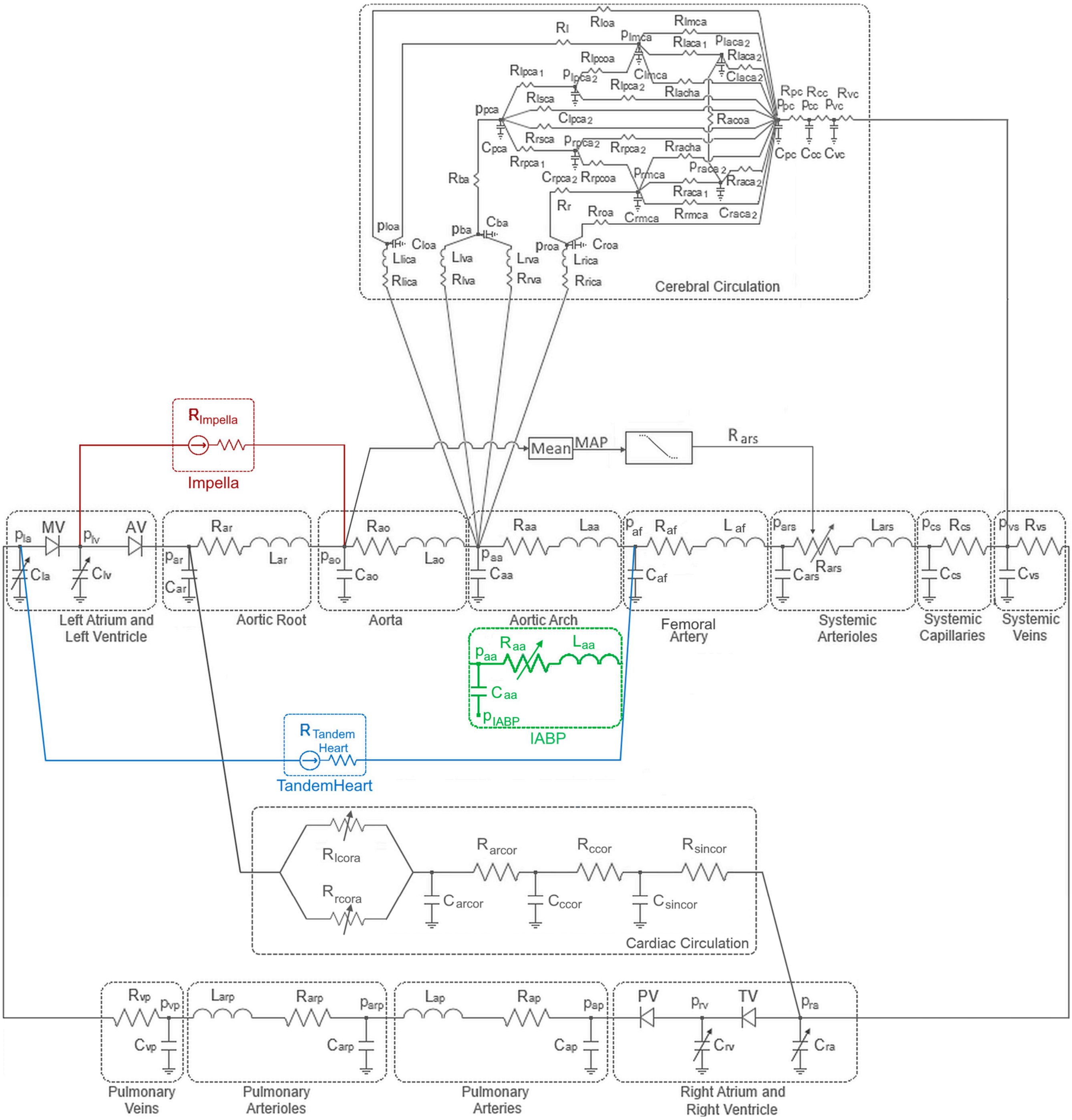

Figure 1. Electric analogue diagram of the cardiovascular system, Impella 2.5, the IABP, and TandemHeart. Numerical models describing Impella 2.5, IABP, and TandemHeart devices were implemented into the cardiovascular system model separately, as given in the electric analogue diagram. Abbreviations used in the figure are given in Table S1.

3. Results

The signal waveforms of the aortic pressure, the left and right coronary arterial and middle cerebral arterial flow rates, and the left ventricular pressure–volume loop in the numerical model simulating CS are given in Figure 2.

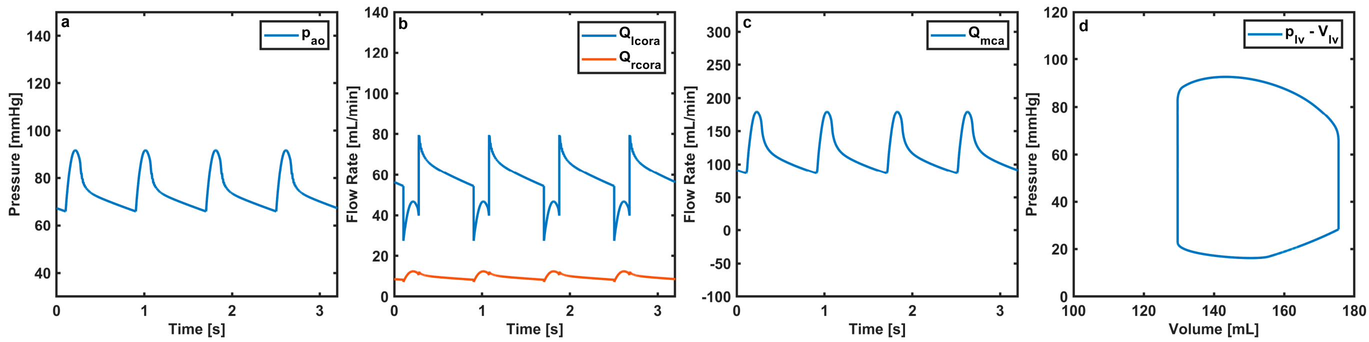

Figure 2. (**a**) Aortic pressure signal waveform, (**b**) left and right coronary arterial signal waveforms, (**c**) middle cerebral arterial signal waveform, and (**d**) left ventricular pressure–volume loop in CS.

The aortic pressure changed between 65 mmHg and 92 mmHg over a cardiac cycle in the numerical model simulating CS. The left coronary arterial flow rate changed between 27 mL/min and 80 mL/min, whereas the right coronary arterial flow rate remained below 20 mL/min over a cardiac cycle in CS. The range of the middle cerebral arterial blood flow rate was between 86 mL/min and 180 mL/min, whereas the end-systolic and end-diastolic volumes were around 103 mL and 175 mL, respectively, in CS. Also, the mean left ventricular filling pressure was around 20 mmHg. The aortic pressure, the left and right coronary arterial and middle cerebral arterial flow rate signal waveforms, and the left ventricular pressure–volume loop in the numerical model simulating IABP support at 0 mmHg, −10 mmHg, and 10 mmHg vacuum pressures and 260 mmHg of drive pressure are given in Figure 3.

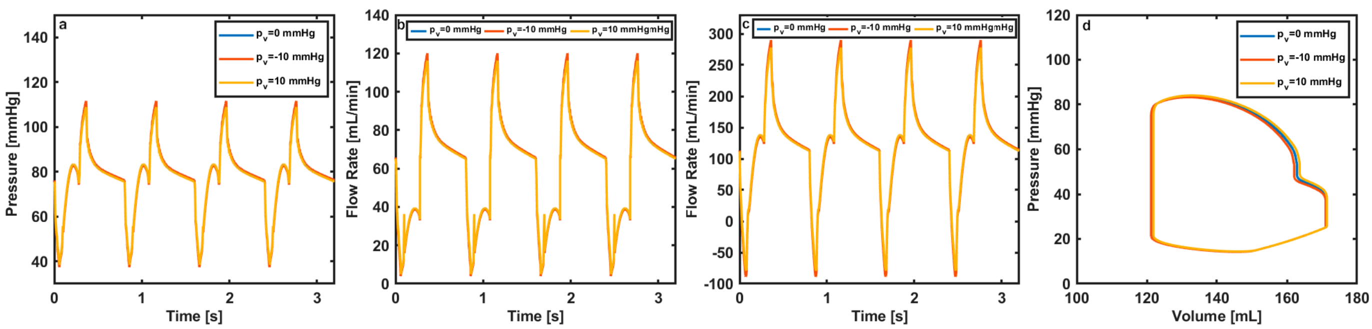

Figure 3. (**a**) Aortic pressure signal waveforms, (**b**) left and right coronary arterial signal waveforms, (**c**) middle cerebral arterial signal waveforms, and (**d**) left ventricle pressure–volume loops at 0 mmHg, −10 mmHg, and 10 mmHg vacuum pressures during IABP support.

The systolic aortic pressure increased to around 110 mmHg, whereas the diastolic aortic pressure decreased to 40 mmHg during IABP support. The left coronary arterial blood flow rate decreased to around 5 mL/min at the end of the diastole and increased to 120 mL/min at the peak systole during IABP support. There was a reverse flow through the middle cerebral artery over the diastolic phase, whereas the maximal middle cerebral arterial blood flow rate increased to 300 mL/min during IABP support. IABP support reduced both the end-systolic and end-diastolic volumes. Varying IABP vacuum pressures slightly affected the aortic pressure, coronary and middle cerebral arterial blood flow rates, and ventricular pressure–volume loop. The aortic pressure, the left and right coronary arterial and middle cerebral arterial flow rate signal waveforms, and the left ventricular pressure–volume loop in the numerical model simulating Impella 2.5 support at 40,000 rpm, 45,000 rpm, and 50,000 rpm operating speeds are given in Figure 4.

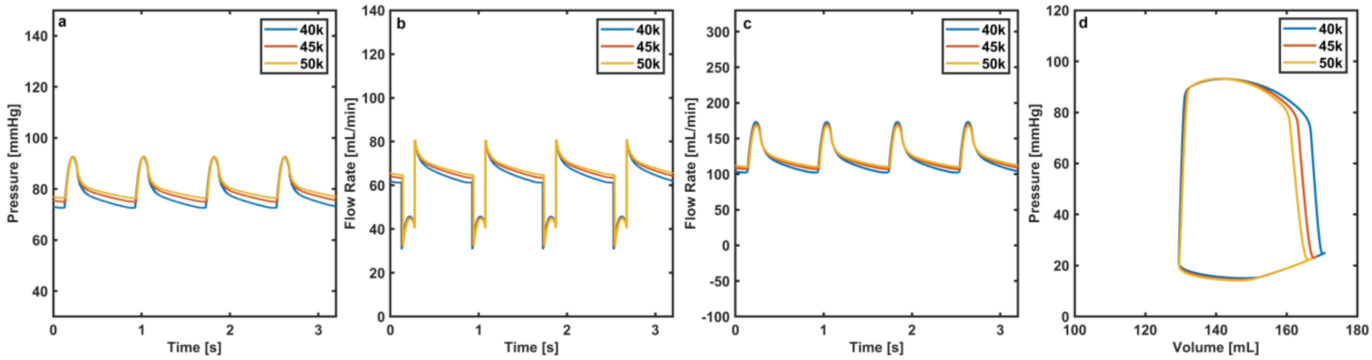

Figure 4. (**a**) Aortic pressure signal waveforms, (**b**) left and right coronary arterial signal waveforms, (**c**) middle cerebral arterial signal waveforms, and (**d**) left ventricular pressure–volume loops at 40,000 (40 k) rpm, 45,000 (45 k) rpm, and 50,000 (50 k) rpm mmHg operating speeds during Impella 2.5 support.

Increasing the Impella 2.5 operating speed increased the diastolic aortic pressure, whereas the systolic aortic pressure changed slightly with Impella 2.5 support. Although the amplitude of the left coronary arterial blood flow rate signal was similar to each other for the simulated Impella 2.5 operating speed, the diastolic blood flow rate increased with the increasing pump speed. The middle cerebral arterial blood flow rate over the diastolic phase also increased with the increasing Impella 2.5 speed. Although the left ventricular end-systolic volume was similar at the simulated Impella operating speed, the left ventricular end-diastolic volume decreased with the increasing pump speed. The aortic pressure, the left and right coronary arterial and middle cerebral arterial flow rate signal waveforms, and the left ventricular pressure–volume loop in the numerical model simulating TandemHeart support at 3000 rpm, 4500 rpm, and 6000 rpm operating speeds are given in Figure 5.

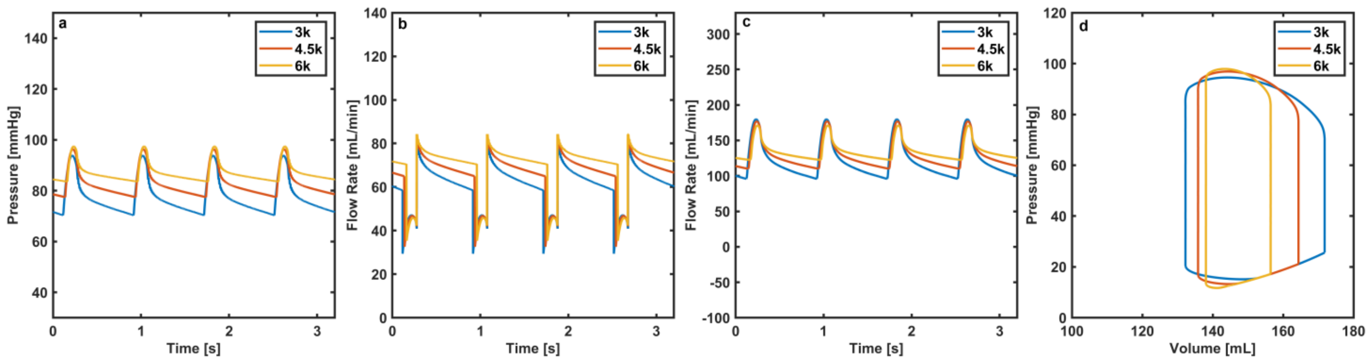

Figure 5. (**a**) Aortic pressure signal waveforms, (**b**) left and right coronary arterial signal waveforms, (**c**) middle cerebral arterial signal waveforms, and (**d**) left ventricle pressure–volume loops at 3000 rpm, 4500 rpm, and 6000 rpm mmHg operating speeds during TandemHeart support.

TandemHeart support increased the overall aortic pressure, and increasing the pump operating speed resulted in relatively high aortic pressure. The increase in the left coronary arterial blood flow rate was more remarkable over the diastolic phase during TandemHeart support. Although the diastolic blood flow rates through the middle cerebral arteries increased with the increasing pump operating speed, there was a decrease in the systolic middle cerebral arterial blood flow rates in the TandemHeart support. The left ventricular pressure volumes were narrowed due to increasing the left ventricular end-diastolic and decreasing the end-systolic volume for the increasing TandemHeart operating speed. The aortic pressure, the left and right coronary arterial and middle cerebral arterial blood flow rate signal waveforms, and the left ventricular pressure–volume loop in the numerical model simulating combined IABP and Impella 2.5 support at 0 mmHg IABP vacuum and 260 mmHg IABP drive pressures and 40,000 rpm, 45,000 rpm, and 50,000 rpm operating speeds are given in Figure 6.

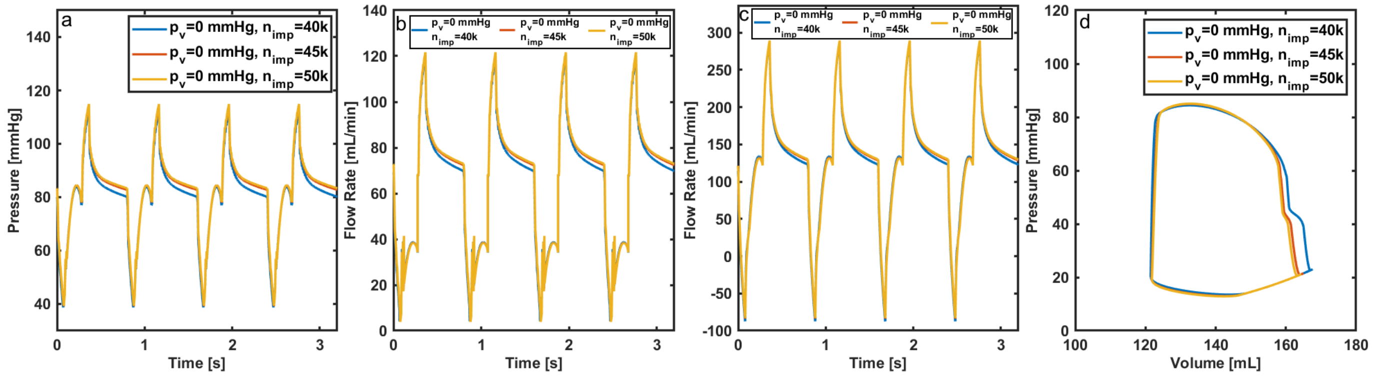

Figure 6. (**a**) Aortic pressure signal waveforms, (**b**) left and right coronary arterial signal waveforms, (**c**) middle cerebral arterial signal waveforms, and (**d**) left ventricle pressure–volume loops at 0 mmHg IABP vacuum and 260 mmHg IABP drive pressures and 40,000 (40 k) rpm, 45,000 (45 k) rpm, and 50,000 (50 k) operating speeds during combined IABP and Impella 2.5 support.

The minimal aortic pressure was around 40 mmHg, whereas the maximal aortic pressure was around 116 mmHg during combined IABP and Impella 2.5 support. The left coronary arterial blood flow rate changed between 120 mL/min and 4 mL/min over a cardiac cycle during combined IABP and Impella 2.5 support. There was a reverse blood flow through the middle cerebral arteries during combined IABP and Impella 2.5 support. Increasing the Impella 2.5 operating speed shifted the left ventricular pressure–volume loops to the right. A comparison of the aortic pressure, the left and right coronary arterial and middle cerebral arterial blood flow rate signal waveforms, and the left ventricular pressure–volume loops in CS, including IABP support at 0 mmHg of vacuum pressure, Impella 2.5 support at a 40,000 rpm operating speed, TandemHeart support at a 3000 rpm operating speed, and combined IABP and Impella 2.5 support at 0 mmHg of vacuum pressure and a 40,000 rpm operating speed, is given in Figure 7.

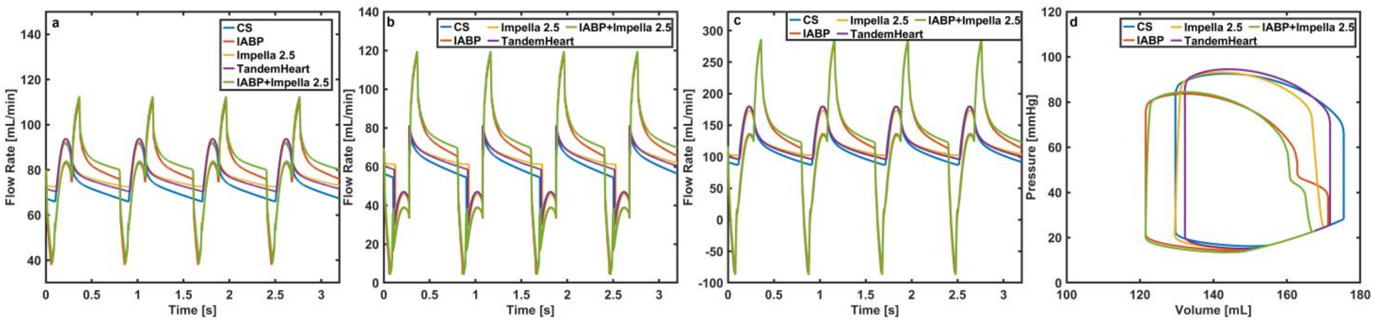

Figure 7. Comparison of (**a**) aortic pressure signal waveforms, (**b**) left and right coronary arterial signal waveforms, (**c**) middle cerebral arterial signal waveforms, and (**d**) left ventricle pressure–volume loops in CS, IABP support at 0 mmHg vacuum pressure, Impella 2.5 support at 40,000 (40 k) rpm operating speed, TandemHeart support at 3000 (3 k) rpm operating speed, and combined IABP and Impella 2.5 support at 0 mmHg vacuum pressure and 40,000 (40 k) rpm operating speed, respectively.

Impella 2.5 support at a 40,000 rpm operating speed and TandemHeart at 3000 rpm operating speed support slightly increased the systolic aortic pressure in comparison to the systolic aortic pressure in CS, whereas there was a remarkable increase in the diastolic aortic pressure under the support of these devices in comparison to the diastolic aortic pressure in CS. The maximal aortic pressure was slightly higher during combined IABP and

Impella 2.5 support than the maximal aortic pressure in the IABP support. An increase in the aortic pressure over the diastolic phase during combined IABP and Impella 2.5 support was noticeable. The effect of continuous unloading of the left ventricle was more noticeable over the diastolic phase in the left coronary and middle cerebral arterial blood flow rate signal waveforms. A reduction in the left ventricular end-diastolic volume was relatively high during IABP support, whereas combining IABP and Impella 2.5 resulted in a relatively low left ventricular end-systolic volume.

Both Impella 2.5 and TandemHeart support reduced the end-systolic volume; however, Impella 2.5 reduced it more than TandemHeart. The overall comparison of the percentage changes in the total blood flow, the left ventricular output, and the mean pump output during IABP support at 0 mmHg, −10 mmHg, and 10 mmHg vacuum pressures and 260 mmHg of drive pressure, Impella 2.5 support at 40,000 rpm, 45,000 rpm, and 50,000 rpm operating speeds, TandemHeart support at 3000 rpm, 4500 rpm, and 6000 rpm operating speeds, combined IABP support at 0 mmHg of vacuum pressure and 260 mmHg of drive pressure, and Impella 2.5 support at 40,000 rpm, 45,000 rpm, and 50,000 rpm operating speeds are given in Figure 8.

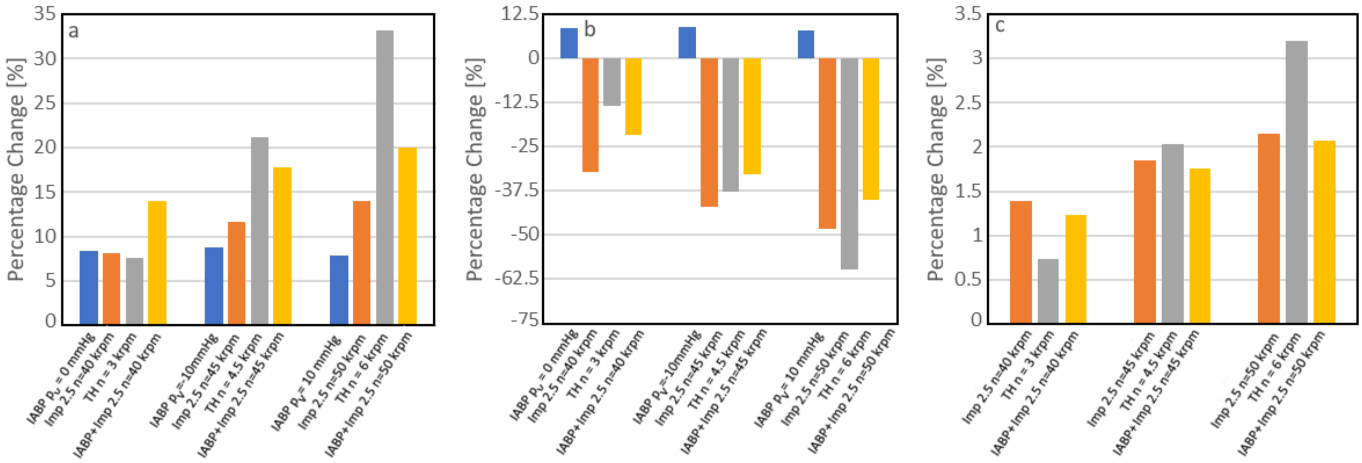

Figure 8. Comparison of percentage changes in (**a**) total blood flow, (**b**) left ventricular output, and (**c**) mean pump output during IABP support at 0 mmHg, −10 mmHg, and 10 mmHg vacuum pressures and 260 mmHg drive pressure, Impella 2.5 support at 40,000 (40 k) rpm, 45,000 (45 k) rpm, and 50,000 (50 k) rpm operating speeds, TandemHeart support at 3000 (3 k) rpm, 4500 (4.5 k) rpm, and 6000 (6 k) rpm operating speeds and combined IABP support at 0 mmHg vacuum pressure and 260 mmHg drive pressure and Impella 2.5 support at 40,000 (40 k) rpm, 45,000 (45 k) rpm, and 50,000 (50 k) rpm operating speeds.

IABP support at 0 mmHg, −10 mmHg, and 10 mmHg vacuum pressures increased the total blood flow rate in the cardiovascular system model by around 8.4%, 8.7%, and 7.8%, respectively. Operating Impella 2.5 at 40,000 rpm, 45,000 rpm, and 50,000 rpm increased the total blood flow rate in the cardiovascular system by around 6.7%, 9.5%, and 11.2%, respectively. TandemHeart support at 3000 rpm, 45,000 rpm, and 6000 rpm increased the total blood flow rate in the cardiovascular system by around 7.6%, 21.2%, and 33.1%, respectively. Combining the IABP and Impella 2.5 support increased the total blood flow further with the increasing Impella 2.5 operating speed. The left ventricular output increased with increased IABP support at the same rates as the total blood flow rate for each vacuum pressure level. The left ventricular output decreased by around 32%, 42%, and 49%, respectively, for the increasing Impella 2.5 operating speed. TandemHeart support at 3000 rpm, 4500 rpm, and 6000 rpm decreased the left ventricular output by around 14%, 38%, and 60%, respectively. Combined IABP and Impella 2.5 support reduced the left ventricular output by around 22%, 33%, and 40%, respectively. Increasing the pump operating speed increased the mean pump output for all the simulated cardiac assist devices. The overall comparison of percentage changes in the mean aortic pressure and diastolic and systolic aortic pressure during IABP support at 0 mmHg, −10 mmHg, and 10 mmHg vacuum pressures and 260 mmHg of drive pressure, Impella 2.5 support at 40,000 rpm, 45,000 rpm, and 50,000 rpm operating speeds, TandemHeart support at 3000 rpm, 4500 rpm, and 6000 rpm operating speeds, combined IABP support at 0 mmHg of vacuum pressure and 260 mmHg of drive pressure, and Impella 2.5 support at 40,000 rpm, 45,000 rpm, and 50,000 rpm operating speeds are given in Figure 9.

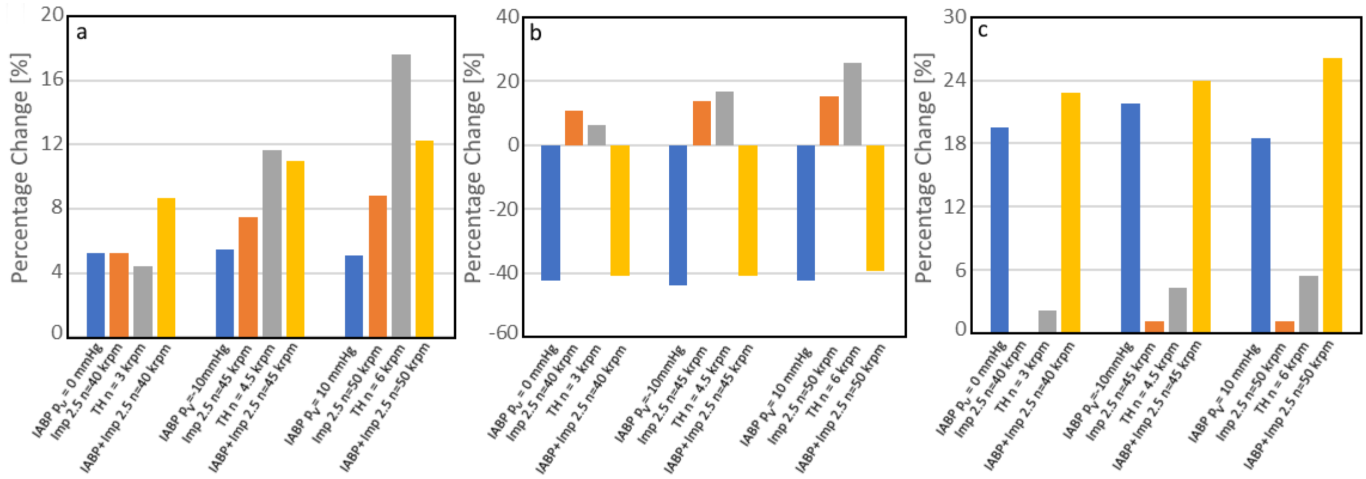

Figure 9. Comparison of percentage changes in (**a**) mean aortic pressure, (**b**) diastolic aortic pressure, and (**c**) systolic aortic pressure during IABP support at 0 mmHg, −10 mmHg, and 10 mmHg vacuum pressures and 260 mmHg drive pressure, Impella 2.5 support at 40,000 (40 k) rpm, 45,000 (45 k) rpm, and 50,000 (50 k) rpm operating speeds, TandemHeart support at 3000 (3 k) rpm, 4500 (4.5 k) rpm, and 6000 (6 k) rpm operating speeds, combined IABP support at 0 mmHg vacuum pressure and 260 mmHg drive pressure, and Impella 2.5 support at 40,000 (40 k) rpm, 45,000 (45 k) rpm, and 50,000 (50 k) rpm operating speeds.

The mean aortic pressure increased by around 5.2% in the IABP support, with 0 mmHg of vacuum pressure and Impella 2.5 support at a 40,000 rpm operating speed, whereas TandemHeart support at a 3000 rpm speed increased the mean aortic pressure by 4.4%. Combined IABP at 0 mmHg and Impella 2.5 operating at 40,000 rpm increased the mean aortic pressure by 8.7%. Although there were minimal changes in the mean aortic pressure for the varying vacuum pressures, increasing the TandemHeart and Impella 2.5 speeds increased the mean aortic pressure. IABP support and combined IABP and Impella 2.5 support decreased the end-diastolic aortic pressure by around 40%. On the other hand, Impella 2.5 and TandemHeart support increased the end-diastolic aortic pressure for the increasing pump operating speed. Impella 2.5 support had minimal effect on the end-systolic aortic pressure, whereas the effect of TandemHeart support on the end-systolic aortic pressure was more noticeable. On the other hand, IABP support and combined IABP and Impella 2.5 support had a profound effect on the end-systolic aortic pressure, as both support modes increased it by more than 18% for the simulated vacuum pressures and operating speeds. The overall comparison of the percentage changes in the left ventricular pressure–volume loop area and the left ventricular end-diastolic and end-systolic volume during IABP support at 0 mmHg, −10 mmHg, and 10 mmHg vacuum pressures and 260 mmHg of drive pressure, Impella 2.5 support at 40,000 rpm, 45,000 rpm, and 50,000 rpm operating speeds, TandemHeart support at 3000 rpm, 4500 rpm, and 6000 rpm operating speeds, combined IABP support at 0 mmHg of vacuum pressure and 260 mmHg of drive pressure, and Impella 2.5 support at 40,000 rpm, 45,000 rpm, and 50,000 rpm operating speeds are given in Figure 10.

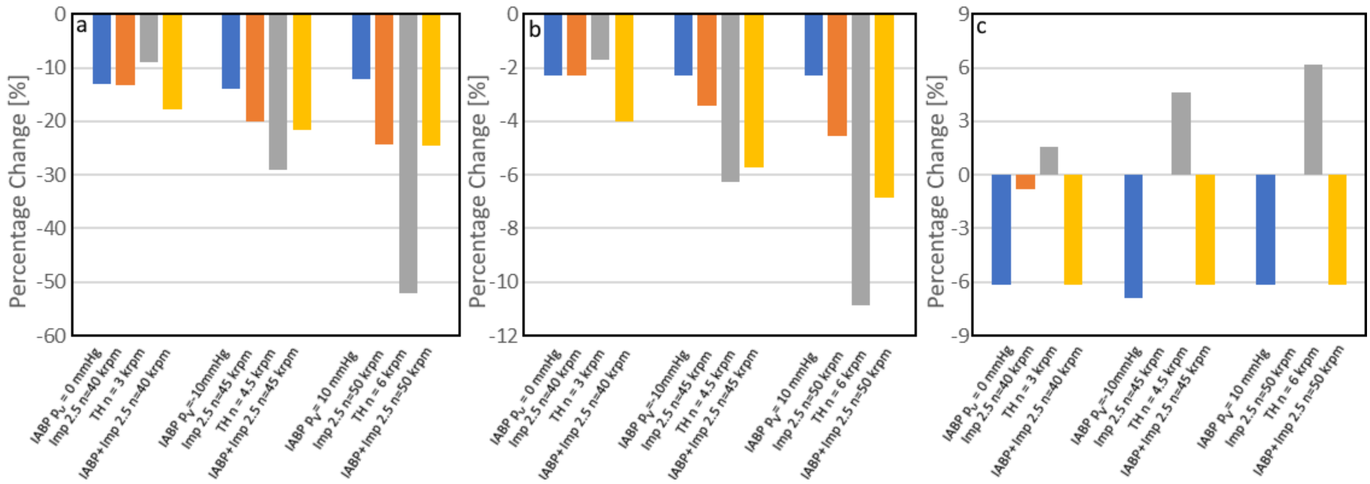

Figure 10. Comparison of percentage changes in (**a**) left ventricular pressure–volume loop area, (**b**) left ventricular end-diastolic volume, and (**c**) left ventricular end-systolic volume during IABP support at 0 mmHg, −10 mmHg, and 10 mmHg vacuum pressures and 260 mmHg drive pressure, Impella 2.5 support at 40,000 (40 k) rpm, 45,000 (45 k) rpm, and 50,000 (50 k) rpm operating speeds, TandemHeart support at 3000 (3 k) rpm, 4500 (4.5 k) rpm, and 6000 (6 k) rpm operating speeds, combined IABP support at 0 mmHg vacuum pressure and 260 mmHg drive pressure, and Impella 2.5 support at 40,000 (40 k) rpm, 45,000 (45 k) rpm, and 50,000 (50 k) rpm operating speeds.

The decrease in the left ventricular pressure–volume loop area under IABP support was more than 10%. Moreover, −10 mmHg of vacuum pressure in the IABP resulted in a relatively high decrease in the left ventricular pressure–volume loop area during IABP support, and increasing the pump operating during Impella 2.5 and TandemHeart support reduced the left ventricular pressure–volume loop area. Combined IABP and Impella 2.5 support reduced the left ventricular pressure–volume loop area by increasing the Impella 2.5 operating speed. However, the decrease in the left ventricular pressure–volume loop area was similar under Impella 2.5 support and combined IABP and Impella 2.5 support at a 50,000 rpm pump operating speed. The decrease in the left ventricular end-diastolic volume was around 2% for the simulated vacuum pressure for the IABP support. Increasing pump operating speeds decreased the left ventricular end-diastolic volume for Impella 2.5, TandemHeart, and combined IABP and Impella 2.5 support. IABP support and combined IABP and Impella support decreased the left ventricular end-systolic volume by around 6% for the simulated configurations. The left ventricular end-systolic volume slightly decreased under Impella 2.5 support at 40,000 rpm, whereas increasing the pump operating speed did not decrease the end-systolic left ventricular volume. TandemHeart support increased the left ventricular end-systolic volume. An overall comparison of the percentage changes in the total coronary arterial and cerebral blood flow rates during IABP support at 0 mmHg, −10 mmHg, and 10 mmHg vacuum pressures and 260 mmHg of drive pressure, Impella 2.5 support at 40,000 rpm, 45,000 rpm, and 50,000 rpm operating speeds, TandemHeart support at 3000 rpm, 4500 rpm, and 6000 rpm operating speeds, combined IABP support at 0 mmHg of vacuum pressure and 260 mmHg of drive pressure, and Impella 2.5 support at 40,000 rpm, 45,000 rpm, and 50,000 rpm operating speeds are given in Figure 11.

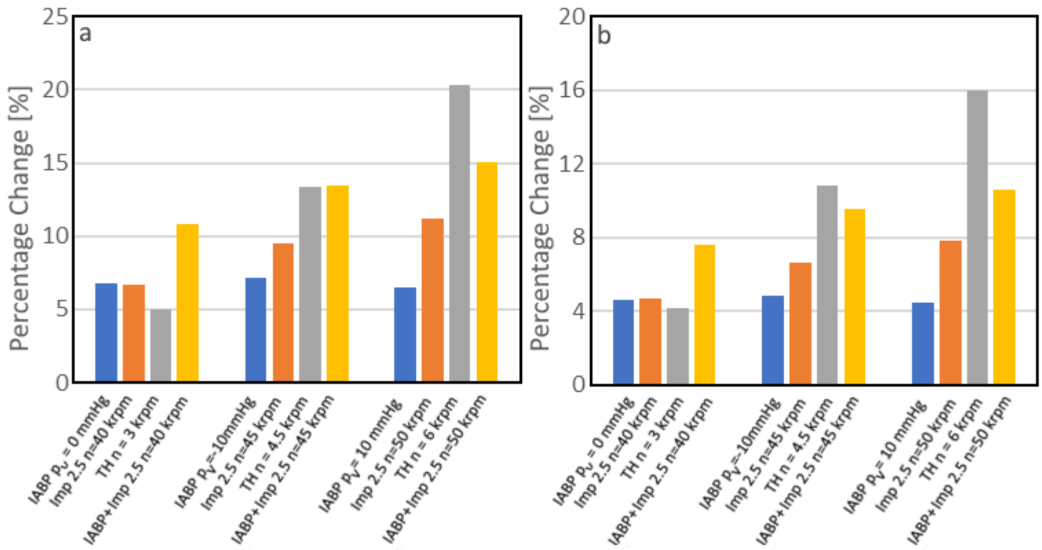

Figure 11. Comparison of percentage changes in (**a**) total coronary blood flow rate, (**b**) total cerebral blood flow rate during IABP support at 0 mmHg, −10 mmHg, and 10 mmHg vacuum pressures and 260 mmHg drive pressure, Impella 2.5 support at 40,000 (40 k) rpm, 45,000 (45 k) rpm, and 50,000 (50 k) rpm operating speeds, TandemHeart support at 3000 (3 k) rpm, 4500 (4.5 k) rpm, and 6000 (6 k) rpm operating speeds, combined IABP support at 0 mmHg vacuum pressure and 260 mmHg drive pressure, and Impella 2.5 support at 40,000 (40 k) rpm, 45,000 (45 k) rpm, and 50,000 (50 k) rpm operating speeds.

The total coronary blood flow rate increased by around 7% in the IABP support. Impella 2.5 support increased the total coronary blood flow rate by around 7% at a 40,000 rpm speed, and increasing the pump operating speed increased the coronary blood flow rate as well. TandemHeart support at 3000 rpm increased the total coronary blood flow rate by around 5%, and increasing the pump operating speed increased the coronary blood flow rate too. Combined IABP support at 0 mmHg of vacuum pressure and Impella 2.5 support at a 40,000 rpm pump operating speed increased the total coronary blood flow rate by around 14%. Again, increasing the pump operating speed increased the total coronary blood flow rate. The total cerebral blood flow rate increased by around 5% in the IABP support. Impella 2.5 support increased the total cerebral blood flow rate by around 5% at a 40,000 rpm speed, and increasing the pump operating speed increased the cerebral blood flow rate as well. TandemHeart support at 3000 rpm increased the total cerebral blood flow rate by around 4%, and increasing the pump operating speed increased the cerebral blood flow rate too. Combined IABP support at 0 mmHg of vacuum pressure and Impella 2.5 support at a 40,000 rpm pump operating speed increased the total cerebral blood flow rate by around 7.5%. Again, increasing the pump operating speed increased the total cerebral blood flow rate. The overall comparison of percentage changes in the mean left ventricular filling pressure during IABP support at 0 mmHg, −10 mmHg, and 10 mmHg vacuum pressures and 260 mmHg of drive pressure, Impella 2.5 support at 40,000 rpm, 45,000 rpm, and 50,000 rpm operating speeds, TandemHeart support at 3000 rpm, 4500 rpm, and 6000 rpm operating speeds, combined IABP support at 0 mmHg of vacuum pressure and 260 mmHg of drive pressure, and Impella 2.5 support at 40,000 rpm, 45,000 rpm, and 50,000 rpm operating speeds are given in Figure 12.

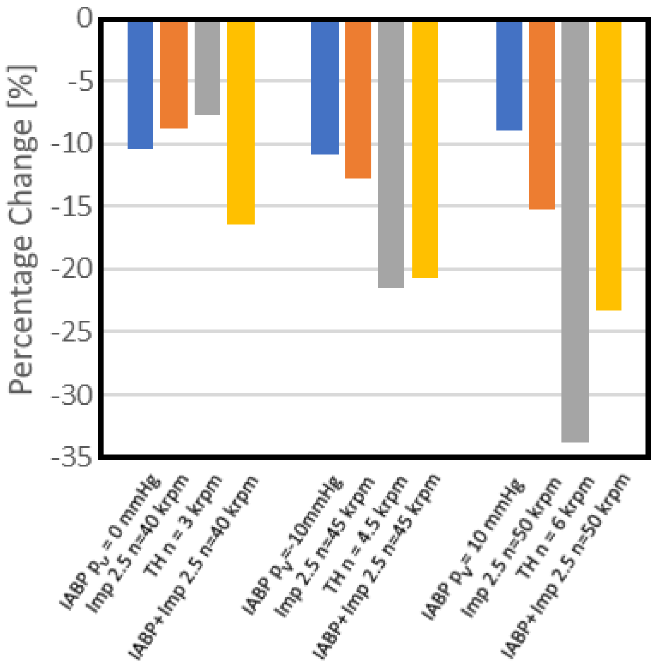

Figure 12. Comparison of percentage changes in mean left ventricular filling pressure during IABP support at 0 mmHg, −10 mmHg, and 10 mmHg vacuum pressures and 260 mmHg drive pressure, Impella 2.5 support at 40,000 (40 k) rpm, 45,000 (45 k) rpm, and 50,000 (50 k) rpm operating speeds, TandemHeart support at 3000 (3 k) rpm, 4500 (4.5 k) rpm, and 6000 (6 k) rpm operating speeds, combined IABP support at 0 mmHg vacuum pressure and 260 mmHg drive pressure, and Impella 2.5 support at 40,000 (40 k) rpm, 45,000 (40 k) rpm, and 50,000 (50 k) rpm operating speeds.

The mean left ventricular filling pressure decreased by 10% during IABP support at 0 mmHg of vacuum pressure, whereas −10 mmHg of vacuum pressure decreased the mean left ventricular filling pressure slightly more. Impella 2.5 support at a 40,000 rpm operating speed decreased the mean left ventricular filling pressure by around 8.5%, whereas TandemHeart operating at 3000 rpm decreased the mean left ventricular filling pressure by around 7.5%. Increasing the pump operating speed during Impella 2.5 and TandemHeart support decreased the mean left ventricular filling pressures further. However, the decrease rate in the mean left ventricular filling pressure was higher in the TandemHeart support. Combined IABP and Impella 2.5 support also had a profound effect on the left ventricular filling pressure. However, at a 6000 rpm operating speed, TandemHeart reduced the mean left ventricular filling pressure by almost 35%, whereas combined IABP and Impella 2.5 support, IABP operating at 0 mmHg of vacuum pressure, and Impella operating at a 50,000 rpm speed decreased the mean left ventricular filling pressure by less than 25%.

4. Discussion

In this study, haemodynamic variables and left ventricular unloading during IABP, Impella 2.5, TandemHeart, and combined IABP and Impella 2.5 support in CS were evaluated using numerical simulations. IABP support at 0 mmHg of vacuum pressure, Impella 2.5 support at a 40,000 rpm operating speed, and TandemHeart support at a 3000 rpm operating speed resulted in an average blood flow rate of around 3.72 L/min in the cardiovascular system model. Increasing the operating speed of the continuous-flow devices or combining IABP and Impella 2.5 yielded different average blood flow rates in the cardiovascular system model. The numerical simulations showed that at similar average blood flow rates in the cardiovascular system, there are minimal differences in the mean aortic pressure during IABP, Impella 2.5, and TandemHeart support. A similar mean aortic pressure at a 3.5 L/min blood flow rate during both IABP and TandemHeart support in CS patients has been reported [20]. The simulation results showed that the mean aortic pressure during combined IABP and Impella 2.5 support was relatively high. However, the blood flow rate also increases with operating IABP and Impella 2.5 together.

The left ventricular pressure–volume loop area is correlated with myocardial oxygen consumption [31]. IABP support at 0 mmHg of vacuum pressure and Impella 2.5 support at a 40,000 rpm operating speed again reduced the left ventricular pressure–volume loop area by 13%, whereas TandemHeart support at 3000 rpm reduced the left ventricular pressure–volume loop area by 9%. Combining IABP and Impella 2.5 at 40,000 rpm decreased the left ventricular pressure–volume loop area further. However, it should be noted that the reduction in the left ventricular pressure–volume loop area in the TandemHeart support is higher than the IABP, Impella 2.5, and combined IABP and Impella 2.5 support. Therefore, it can be concluded that TandemHeart may provide better support in terms of cardiac energetics.

IABP support reduced both the end-systolic and end-diastolic volumes in the left ventricle, whereas partial Impella 2.5 support reduced only the end-diastolic volume whilst increasing the systolic pressure in the left ventricle. Again, this is an expected outcome when IABP and Impella 2.5 devices are compared [32].

IABP and Impella 2.5 support provide similar coronary blood flow rates at 0 mmHg of vacuum pressure and a 40,000 rpm operating speed. Increasing the operating speed of Impella 2.5 resulted in better coronary perfusion in comparison to IABP support. Although the coronary blood flow rate during TandemHeart support is relatively low, it increases profoundly with the increasing TandemHeart operating speed. Combining IABP and Impella 2.5 and increasing the operating speed of Impella 2.5 improved coronary perfusion as well. Similar outcomes were obtained for the total cerebral blood flow rates during IABP, Impella 2.5, TandemHeart, and combined IABP and Impella 2.5 support. Experimental data show that Impella 2.5 provides higher carotid arterial blood flow rates, whereas it is possible to improve it by combining IABP and Impella 2.5 [33], confirming the simulation results. However, it should be noted that there was a reverse blood flow rate through the middle cerebral arteries during IABP support because the amplitude of the aortic pressure increases due to the decreased end-diastolic and increased end-systolic pressure. Again, such an outcome may be expected during IABP support [34,35]. However, combining IABP and Impella 2.5 did not avoid the reversal of the blood flow. Therefore, the use of TandemHeart may be more beneficial for cerebral perfusion in CS.

IABP support reduced the mean left ventricular filling pressure by around 10% for the simulated configurations. Although the reduction in the mean left ventricular filling pressure on Impella 2.5 support was higher with increasing the pump operating speed, TandemHeart and combined IABP and Impella 2.5 support had a more profound effect on this parameter. Therefore, TandemHeart and combined IABP and Impella 2.5 support may be more beneficial in reducing the left ventricular filling pressure in CS.

The IABP-SHOCK II trial suggested that IABP support did not lead to a reduction in 12-month mortality rates among patients with cardiogenic shock [15]; however, the IABP continues to be the predominant device employed for this patient cohort [14]. The simulation results in this study provide insights into the haemodynamic and energetic outcomes in CS during MCS. IABP support increases the mean aortic pressure, systolic aortic pressure, and coronary and cerebral perfusion, whereas it reduces the left ventricular pressure–volume loop area. However, it would appear that the benefits of IABP support on the hemodynamic and energetic outcomes are relatively low when compared to Impella 2.5 and TandemHeart. Combining IABP and Impella 2.5 profoundly affects the mean aortic pressure and coronary and cerebral perfusion at a relatively high operating speed for Impella 2.5. However, the reversal of cerebral blood flow may be a limitation. On the other hand, TandemHeart had a profound effect on the mean aortic pressure, coronary and cerebral perfusion, and myocardial oxygen consumption, especially at a relatively high operating speed.

5. Conclusions

Based on the simulation results, TandemHeart may have a more beneficial effect on haemodynamics and left ventricular energetics in comparison to IABP and Impella 2.5. However, the downside of using TandemHeart is the need for expertise in trans-septal catheterisation, which may not be readily available. The potential for residual atrial septal defect is another issue to consider. In view of these considerations, a combined use of IABP and Impella 2.5 may prove to be an appropriate alternative for short-term support, with a view to upgrading to more advanced devices if required.

Supplementary Materials: The following supporting information can be downloaded at: https://www.mdpi.com/article/10.3390/math11163606/s1, Table S1: Abbreviations used in the cardiovascular system model and equations.

Author Contributions: Conceptualization, C.D.L. and S.B.; methodology, R.A.; software, R.A.; validation, R.A., B.D.L., M.C., C.D.L. and S.B.; formal analysis, B.D.L., M.C., C.D.L. and S.B.; investigation, R.A.; writing—original draft preparation, R.A.; writing—review and editing, B.D.L., M.C., C.D.L. and S.B.; supervision, C.D.L. and S.B.; project administration, S.B. All authors have read and agreed to the published version of the manuscript.

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