In-vitro Evaluation of Physiological Controller Response of Rotary Blood Pumps to Changes in Patient State

Jo P Pauls, Shaun D Gregory, Michael Stevens and Geoff Tansley

Abstract— **Rotary blood pumps (RBPs) have a low sensitivity to preload changes when run at constant speed, which can lead to harmful ventricular suction events. Therefore a control mechanism is needed to adjust RBP speed in response to patient demand, but an appropriate response time for physiological control strategies to these changes in patient demand has not been determined. This paper aims to evaluate the response of a simulated healthy heart with those of different RBP control techniques during exercise simulations and a Valsalva manoeuver. A mock circulation loop was used to simulate the response of a healthy heart to these changes in patient state. The generated data was compared with a simulated RBP (VentrAssist) supported left heart failure condition. A range of control techniques including constant speed, proportional integral (PI) (active control) and a compliant inflow cannula (passive control) were used to achieve restored haemodynamics and evaluate controller response time.**

Controllers that responded faster (active control) or slower (active control and constant speed mode) than the native heart's response led to ventricular suction. Active control systems can respond both faster or slower than the heart depending on the controller gains. A control system that responded similar to the native heart was able to prevent ventricular suction. This study concluded that a physiological control system should mimic the response of the native heart to changes in patient state in order to prevent ventricular suction.

I. Introduction

Rotary blood pumps (RBPs) can be used for mechanical circulatory support during bridge to transplant, bridge to recovery, or destination therapy for patients unsuitable for transplantation [1]. The healthy heart responds to an increase in preload with an increase in contraction force (Frank-Starling response) in order to balance outputs of both ventricles. The preload sensitivity of RBPs, however, is significantly less than that of the native heart [2]. This preload insensitivity means that pump flow, when RBPs run in a constant speed mode, cannot passively change sufficiently in response to frequent changes in preload during exercise, straining (Valsalva manoeuver) or in response to illness [3]–[5]. Changes in pump flow are necessary to avoid adverse events such as ventricular suction or pulmonary oedema. Ventricular suction may lead to haemolysis, heart tissue damage near the pump inlet, right ventricular dysfunction or release of ventricular thrombus and subsequent stroke [6], [7]. Accordingly, various clinical studies have reported the importance of physiological controller development to prevent suction events [8], [9].

Various physiological control systems for RBPs have been developed, but an appropriate controller response time has not been determined. Salamonsen et al. states that a physiological control system should respond within 8 heartbeats (approximately 10 s or less) [10], but there is no mention of a controller responding too quickly.

A physiological control system should mimic the response of the native heart to changes in preload during varying patient states. This paper compares the response of a simulated healthy heart with those of different RBP control techniques during exercise simulations and a Valsalva manoeuver. It characterizes the requirements of a physiological controller for RBPs to match the healthy autonomous system.

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II. Methods

A. Mock Circulation Loop

A physical five element Windkessel mock circulation loop (MCL) including systemic and pulmonary circulations was used for this study [11], [12]. Atrial and ventricular chambers were represented by clear, vertical polyvinyl chloride pipes with tee sections connecting the inflow, outflow and heart chamber. Ventricular systole was controlled through a series of electropneumatic regulators (ITV2030-012BS5, SMC Pneumatics, Tokyo, Japan) and 3/2 way solenoid valves (VT325-035DLS, SMC Pneumatics, Tokyo, Japan) to provide passively filled heart chambers and variable contractility, heart rate and systolic time. A Starling response was implemented in both left and right ventricles which actively controlled ventricular contractility (through electropneumatic regulator supply current) based on ventricular preload [13]. Mechanical check valves were used to simulate the mitral, aortic, tricuspid and pulmonary valves. Four independent Windkessel chambers were employed to simulate lumped systemic and pulmonary arterial and venous compliance. Socket valves (VMP025.03X.71, AKO Alb. Klein Ohio LLC, USA) allowed easy manipulation of systemic and pulmonary vascular resistance. The working fluid throughout this study was a water/glycerol mixture (60/40% by mass) which, at a room temperature of 22°C, demonstrated similar viscosity

(3.5 mPa.s) and density (1100 kg/m^3) to that of blood at 37°C.

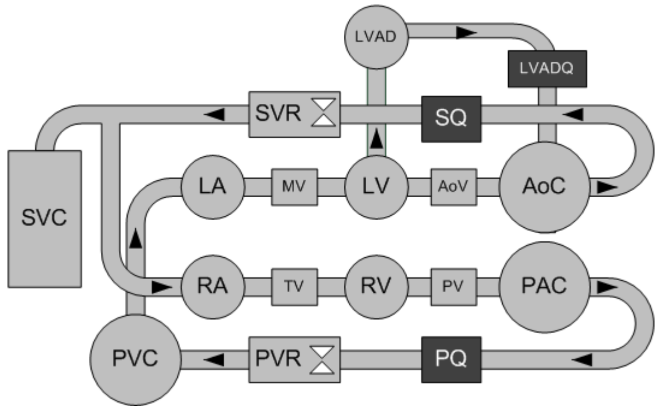

Figure 1. Schematic of the MCL setup. LA - left atrium, MV - mitral valve, LV - left ventricle, AoV - aortic valve, AoC - aortic compliance chamber, SVR - systemic vascular resistance valve, SVC - systemic venous compliance chamber, RA - right atrium, TV - tricuspid valve, RV - right ventricle, PV - pulmonary valve, PAC - pulmonary arterial compliance chamber, PVR - pulmonary vascular resistance valve, PVC - pulmonary venous compliance chamber, LVAD – left ventricular assist device, LVADQ - left ventricular assist device flow

B. *Mock Circulation Loop Test Beds*

Test beds for simulating changes in patient state (exercise and the Valsalva manoeuver) were implemented in the MCL using Simulink/Matlab (Matlab 2009a, MathWorks, Natick, MA, USA) and dSPACE ControlDesk (DS1104, dSPACE, Wixom, MI, USA). Exercise was simulated by increasing heart rate (HR) and both left/right ventricular contractility while decreasing systemic and pulmonary vascular resistance and shifting fluid from the systemic venous compliance chamber into the heart via the right atrium [14]. The simulated exercise intensity was dependent on the magnitude of changes in HR, ventricular contractility, systemic and pulmonary vascular resistance and the amount of shifted fluid. Transfer functions were implemented in the exercise test bed for each variable to be able to simulate the time response of the native heart to exercise, which was investigated by our group in a previous study. A Valsalva manoeuver was simulated by increasing the air pressure in the pulmonary vascular compliance chamber (‘breathing in’) using electropneumatic regulators (ITV2030-012BS5, SMC Pneumatics, Tokyo, Japan) and holding that level for a set amount of time while increasing PVR.

C. *In-Vitro Experiments*

The MCL with the implemented test beds was initially used to simulate the response of a healthy heart during exercise and a Valsalva manoeuver. The MCL was then configured to represent a medically treated left heart failure condition without LVAD support. Haemodynamics were restored by connecting a VentrAssist (Ventracor Ltd., Sydney, Australia) RBP to the left ventricle for inflow and the aorta for outflow (see Figure 1).

Both active and passive control systems were used in this study to evaluate the response of various physiological controllers. Two Frank-Starling like controllers by Stevens et al. [15] and Salamonsen et al. [10] were tested in this study as active control systems. The control strategy proposed by Stevens et al. replicated the preload sensitivity of the native heart through automatic adjustments in pump speed, and thus pump flow rate, based on left ventricular end diastolic pressure (LVEDP) as the feedback variable. The controller described in Salamonsen et al. employed the same technique, however used pump flow pulsatility as a controller surrogate for LVEDP. A compliant inflow cannula developed by Gregory et al. [16] was evaluated as a passive control system. The compliant inflow cannula showed to passively increase resistance of a RBP circuit and therefore reduced RBP outflow when preload decreased, thus preventing ventricular suction. To validate the need for a physiological control system the RBP run in constant speed mode was also tested.
The ability of the control systems to automatically alter RBP flow rates similarly to the native heart while preventing ventricular suction events was then evaluated during exercise and the Valsalva manoeuver. To compare the control systems, a time constant (τ) was calculated as the time it took the systemic flow rate to reach 63.2% of its final value during exercise simulation.

D. *Data Acquisition*

Haemodynamic and LVAD parameters were captured at 100 Hz using a dSPACE acquisition system (DS1104, dSPACE, Wixom, MI, USA). Systemic and pulmonary flow rates were recorded using magnetic flow meters (IFC010, KROHNE, Duisburg, Germany) while LVAD outlet flow rate was recorded with a clamp-on ultrasonic flow meter (TS410-10PXL, Transonic Systems, Ithaca, NY, USA). Circulatory and LVAD pressures were recorded using silicon-based transducers (PX181B-015C5V, Omega Engineering, Stamford, CT, USA) while left and right ventricular volumes were recorded using magnetostrictive level sensors (IK1A, GEFRAN, Provaglio d’Iseo, Italy).

III. RESULTS

A. *Exercise Simulations*

All control systems and the RBP run in constant speed mode were able to achieve a systemic flow rate comparable to that of the simulated healthy heart (Figure 2).

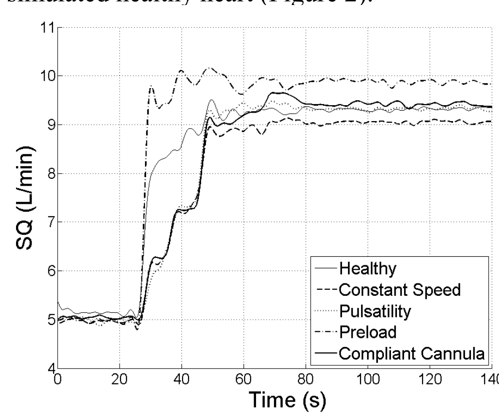

Figure 2. Systemic flow rate (SQ) vs. Time during exercise simulation.

The active controller using preload as a feedback variable showed the fastest response (τ = 3-4 seconds) to exercise compared to the other control systems. The pulsatility controller, the compliant cannula and the pump at constant speed all responded more slowly than the simulated healthy heart (τ = 20 seconds on average).

A comparison of LVAD flow rates (Figure 3) shows similar results in response times to the systemic flow rates. Only the two active control systems increased LVAD flow significantly (from 4 L/min to 8 L/min (preload) or 6.5 L/min (pulsatility)) during exercise simulations by increasing pump speed. LVAD flow increased from 4 L/min to 5.5 L/min with the compliant cannula and the pump run in constant speed mode, due to increased ventricular contractility, lower SVR and the small amount of preload sensitivity of the pump.

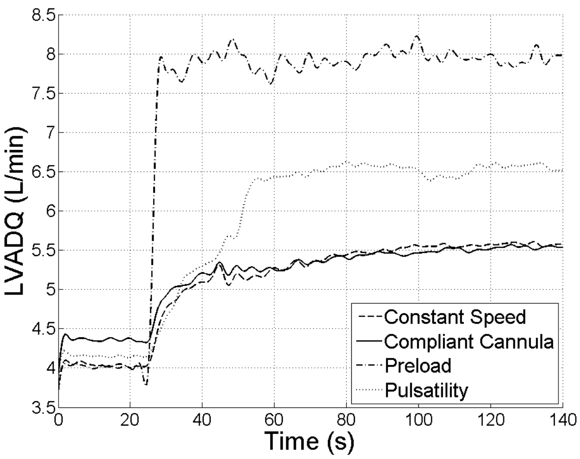

Figure 3. Left ventricular assist device flow (LVADQ) vs. Time during exercise simulation.

B. *Valsalva Manoeuver Simulations*

Figure 4 shows the systemic flow rate comparison during a Valsalva manoeuvre simulation. The preload controller responded the fastest to the Valsalva manoeuvre which led to overshoot of pump speed, and flow rate, in the early response compared to the healthy condition. The pulsatility controller and the pump run in constant speed mode responded more slowly than the healthy heart. The compliant inflow cannula mimicked the response of the native heart most closely and showed similar response times with a lower magnitude in systemic flow changes throughout the Valsalva manoeuver.

When left ventricular volumes (LVV) were compared a rapid increase in LVV was observed in all cases at the start of the Valsalva manoeuvre followed by a decrease in LVV (Figure 5). Ventricular suction events were defined as LVV equal or less than zero. The fast-responding preload controller showed the most rapid decrease in LVV until ventricular suction occurred, observed when LVV reaches the zero level. The slower-responding pulsatility controller and the pump run at constant speed mode showed a slower decrease in LVV and showed ventricular suction later during the Valsalva manoeuvre. The compliant cannula, which responded most similarly to the native heart, prevented ventricular suction throughout the experiment.

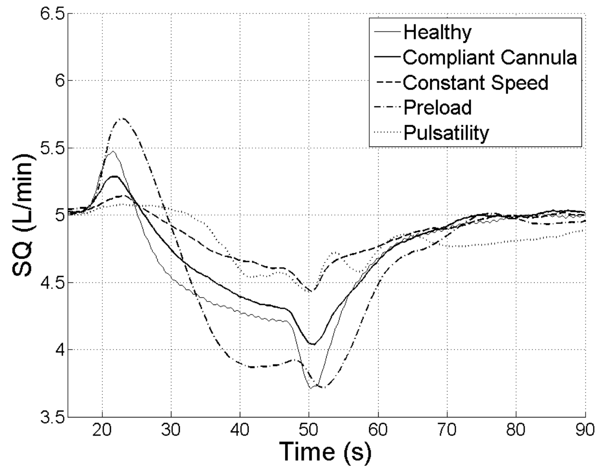

Figure 4. Systemic flow rate (SQ) vs. Time during Valsalva simulation.

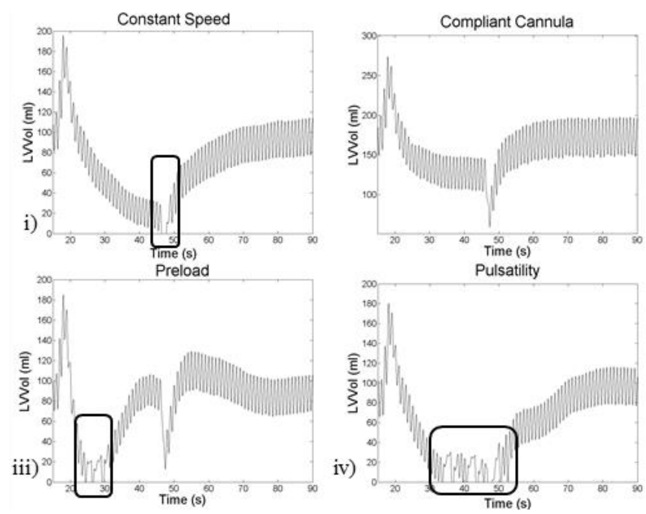

Figure 5. Left ventricular volume (LVV) vs. time for i) constant speed, ii) compliant cannula, iii) preload controller and iv) pulsatility controller. Left ventricular suction events are marked with a black rectangle.

IV. DISCUSSION

In-vitro results of exercise simulations showed that similar systemic flow rates were achieved with all tested control systems when compared to the healthy heart. Only the two active controllers increased LVADQ by increasing pump speed according to a change in demand. The compliant inflow cannula and the pump in constant speed mode were not able to increase LVADQ like the active controllers could. The increase in flow observed in Figure 3 was due to a sudden decrease in systemic vascular resistance and in small part due to the preload sensitivity of the pump. Systemic flow in these two cases was able to reach 9 – 10 L/min during exercise due to the remnant contractility of the failed ventricle forcing ejection through the aortic valve.

This led to an increase in left ventricular stroke work (LVSW) from 0.1 J at rest to 0.7 J during exercise when the RBP was run in constant speed mode or a compliant inflow cannula was used. The preload controller, on the other hand, only increased LVSW from 0.1 J at rest to 0.2 J during exercise. This shows that the left ventricle has to work harder when the compliant cannula is used or the pump is run at a constant speed and thus may not recover as well. Therefore a physiological control system should increase the LVADQ during exercise to accommodate for the higher demand in cardiac output, achieve an increased exercise capacity while unloading the heart to reduce the workload on the failing heart.

The response of the heart and the control systems to a Valsalva manoeuver was generally faster when compared to exercise response times. A comparison of LVV revealed that control systems that respond too quickly (preload controller) or control systems that respond too slowly (pulsatility controller and constant speed mode) were not able to prevent ventricular suction. The control system that mimicked the response of the simulated heart the closest (compliant inflow cannula) was able to prevent suction events.

To accommodate for the different response times of the heart to the various changes in patient states a combination of an active and passive physiological control system might be needed for an optimized control system. This way the compliant inflow cannula could prevent ventricular suction during more rapid changes while the active control system would be able to increase pump flow when needed, e.g. during exercise. The response time of the active control systems might be altered to mimic the response of the healthy heart more by optimizing the controller gains.

V. Conclusion

This study showed that a simulated healthy heart responded slower during exercise and faster during a Valsalva manoeuver. Physiological control systems for RBPs should mimic the response of the healthy heart to changes in patient state in order to avoid adverse events. The compliant cannula responded similar to the healthy heart during the Valsalva manoeuver and was able to prevent ventricular suction. The two active control systems were able to increase LVADQ during exercise, which helped to unload the failing heart and might help during recovery.

Both active and passive control techniques improve RBP response to changes in patient states compared to constant speed mode. A combined, active and passive control approach, where both control systems complement each other, should be investigated in the future and may provide a better quality of life for heart failure patients supported by RBPs.

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