33rd Annual International Conference of the IEEE EMBS
Boston, Massachusetts USA, August 30 - September 3, 2011

Frank-Starling Control of a Left Ventricular Assist Device

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***Abstract*—A physiological control system was developed for a rotary left ventricular assist device (LVAD) in which the target pump flow rate (LVADQ) was set as a function of left atrial pressure (LAP), mimicking the Frank-Starling mechanism. The control strategy was implemented using linear PID control and was evaluated in a pulsatile mock circulation loop using a prototyped centrifugal pump by varying pulmonary vascular resistance to alter venous return. The control strategy automatically varied pump speed (2460 to 1740 to 2700 RPM) in response to a decrease and subsequent increase in venous return. In contrast, a fixed-speed pump caused a simulated ventricular suction event during low venous return and higher ventricular volumes during high venous return. The preload sensitivity was increased from 0.011 L/min/mmHg in fixed speed mode to 0.47L/min/mmHg, a value similar to that of the native healthy heart. The sensitivity varied automatically to maintain the LAP and LVADQ within a predefined zone. This control strategy requires the implantation of a pressure sensor in the left atrium and a flow sensor around the outflow cannula of the LVAD. However, appropriate pressure sensor technology is not yet commercially available and so an alternative measure of preload such as pulsatility of pump signals should be investigated.**

I. Introduction

Rotary left ventricular assist devices (LVADs) can successfully support a failing left ventricle (LV) whilst the patient awaits recovery or transplant, or be used as destination therapy for those deemed unsuitable for transplantation [1]. However, one limiting factor of rotary LVADs is that their preload sensitivity is significantly smaller than that of the native heart, which lies between 0.213 and 0.9 L/min/mmHg [2, 3].

This preload insensitivity means that, when operated at a constant speed, the pump output cannot passively change sufficiently in response to the frequent variations in preload induced by changes in posture, exercise, straining or in response to intercurrent illness. [4-6]. Speed changes are required to avoid hazardous events including ventricular suction, which results in reduced forward flow of blood, hence inducing ischaemia on the heart as well as distal organs, haemolysis, release of ventricular thrombus leading to stroke, tissue damage at VAD inlet and even subsequent right ventricular dysfunction [7-9]. The addition of an active physiological control system is required to automatically match LVAD flow with venous return, thus ensuring appropriate cardiac output at all times whilst avoiding ventricular suction.

A number of physiological control systems for rotary LVADs have already been developed that detect imminent suction by monitoring pump signals (such as current, differential pressure, and power) and operate the pump at the highest speed possible without suction occurring [10, 11]. However, this strategy does not prevent suction during sudden drops in venous return, which may occur as the patient stands up or during straining. It can also be argued that it is unphysiological to maintain a constant ventricular volume, especially the low volume maintained in these suction-avoidance control strategies, as it may disrupt right ventricular function.

The Frank-Starling law is responsible for relating cardiac output to preload, ensuring the outflow from the ventricles matches the venous return. This mechanism is diminished in heart failure due to weaker ventricles. In this work, we propose a control strategy for a rotary LVAD, in which the LVAD flow (LVADQ) is functionally dependent on preload, represented by left atrial pressure (LAP), in order to directly mimic the Frank-Starling mechanism. This strategy measures LAP directly using a pressure sensor, and sets a target LVADQ using characteristic Frank-Starling curves described by Guyton [12]. A linear proportional, integral and derivative (PID) controller is then used to vary the pump speed to control the mean LVADQ, measured using a flow sensor. This strategy was evaluated in a pulsatile mock circulation loop (MCL) using a prototyped rotary pump. Its response to changes in venous return, caused by variations in pulmonary vascular resistance, was compared to that of a LVAD operated at a constant speed.

II. Materials and Methods

A. Mock Circulation Loop and Rotary Pump

A MCL is a mechanical representation of the human heart and circulatory system. The MCL used in this study is a 5-element Windkessel biventricular model of the circulatory system, which uses a regulated supply of compressed air to pump fluid through a simulated cardiovascular circuit [13].

Manuscript received June 20, 2011. This work was supported in part by The Medical Advances Without Animals Trust (MAWA), The Prince Charles Hospital Foundation and the Critical Care Research Group.

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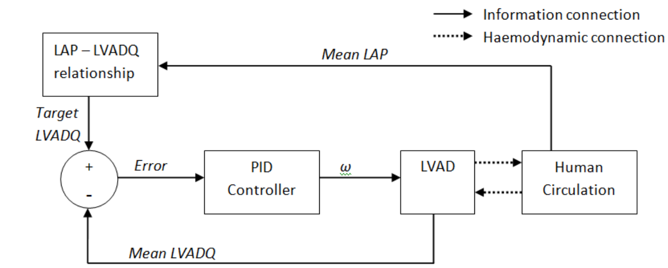

Fig. 1. Block diagram of the LVAD physiological control strategy. ω = pump rotational speed.

This circuit simulates cardiovascular resistance, inertance and compliance in both pulmonary and systemic circulation. Left and right atrial and ventricular pressures, aortic pressure, pulmonary arterial pressure and LVAD inlet/outlet pressures were monitored using silicone based transducers (PX181B-015C5V, OMEGA Engineering, Connecticut, USA). Systemic, pulmonary and LVAD flow rates were monitored using an ultrasonic flow meter (TS410-10PXL, Transonic Systems, NY, USA). All data were recorded using a dSPACE 1103 controller board (Ceanet Pty Ltd, Sydney, Australia). A solution of 40wt% glycerol was used as a surrogate for human blood. The loop was modified so that its ventricles exhibited a native Frank-Starling mechanism.

A prototyped mixed-flow rotary pump with an impeller diameter of 50mm was used to simulate a rotary LVAD. The impeller was driven by a DC motor (AMax 236669, Maxon Motors, Sachseln, Switzerland). The rotational speed of the DC motor was measured using a digital encoder (143306 HEDL, Maxon Motors, Sachseln, Switzerland).

B. Control Strategy

The effect of the Frank-Starling mechanism is that cardiac output (CO) is dependent on preload. Specifically, the CO increases with increasing LAP until it saturates at a maximum [12]. Our control algorithm aims to make the blood flow rate through the pump dependent on preload, replicating the preload sensitivity of the native heart. A block diagram of the complete control system is shown in Figure 1. The LAP is measured directly using a pressure sensor, and a target LVADQ is set based upon a functional relationship between these two variables. A PID controller was used to adjust pump speed to minimise the error between the target and measured LVADQ. The PID controller gains were set manually using the Zeigler-Nicholls tuning method. Both the measured LAP and the measured LVADQ were passed through low pass filters with a cut-off frequency of 0.5Hz to obtain their mean values.

The functional relationship between LVAD flow (LVADQ) and LAP is interpolated from the Frank-Starling characteristic curves described by Guyton [12]. These curves were normalised and a function was fitted using LAB FIT software (Universidade Federal de Campina Grande, Campina Grande, Paraíba, Brazil) giving the function:

$$LVADQ = 7 \times \left(\frac{-5.259+LAP}{(7.124+0.2665LAP^2)} + 0.8698\right) \qquad (1)$$

The sensitivity of these curves is hereby defined as the approximate slope of the linear portion of Equation 1. This relationship was scaled along the pressure axis to produce curves of different sensitivities, which reflects the different inotropic states of the heart (Figure 2). When used with a constant sensitivity (denoted as "Controller On, Autosensitivity Off" in the results), the control strategy should be able to ensure no suction or overfilling during preload changes caused by straining and changes in posture. However, no single sensitivity is appropriate for all scenarios. Changes in a patient's cardiac demand or heart condition may require a different sensitivity in order to avoid suction or overfilling. The control strategy must therefore be able to vary the sensitivity automatically (denoted as "Controller On, Autosensitivity On" in the results). Our control strategy maintains the sensitivity constant until the current values of LVADQ and LAP (herein referred to as the operating point (OP)) move outside a predefined ideal operating zone (IOZ). A proportional plus integral (PI) controller, tuned manually using the Zeigler-Nichols tuning method, is then used to adjust the sensitivity until the OP returns to the IOZ. The IOZ is defined by a minimum and maximum LAP and LVADQ. The IOZ should be set by a clinician at the time of implantation, based on what they believe are safe boundaries for each particular patient. For this study the IOZ was defined by a minimum and maximum LAP of 4 and 8 mmHg, and a minimum and maximum LVADQ of 2 and 5 L/min respectively, in order to avoid suction events and to demonstrate the autosensitivity control.

C. Simulation Protocol

A MCL was used to evaluate the control strategy. Five scenarios were simulated in the MCL: a normal heart, LV failure without an LVAD, LV failure supported by an LVAD operated at fixed speed, and LV failure supported by a LVAD operated using the Frank-Starling control strategy, both with and without the autosensitivity control. It was assumed right heart function remained normal during the LV failure scenarios.

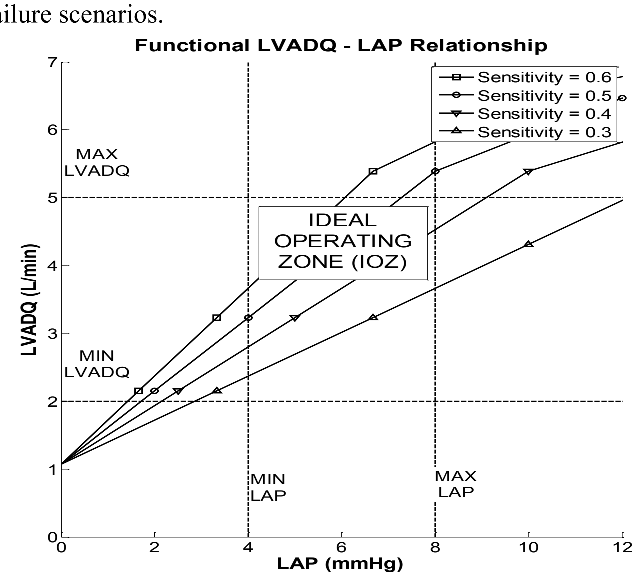

Fig. 2. Functional relationship between LVAD flow rate (LVADQ) and left atrial pressure (LAP), which was used to set a target flow rate for a specific LAP.

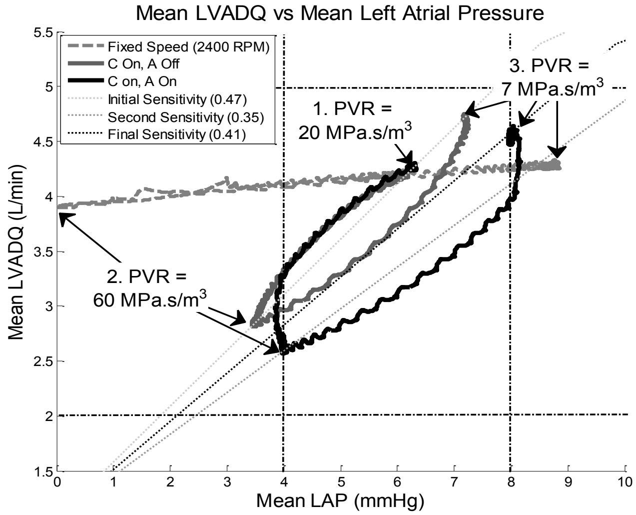

Fig. 3. Mean LVADQ vs Mean LAP over time, showing the preload sensitivity of each of three LVAD control strategies tested in the MCL. C On, A Off = controller on, autosensitivity control off. C On, A On = controller on, autosensitivity control on. PVR = pulmonary vascular resistance

The simulation protocol was as follows: First, the MCL was primed with a mean vascular pressure of 8.5mmHg. The ventricles of the loop were then started and the appropriate heart condition was set by adjusting the ventricular contractilities and vascular resistances accordingly. The initial haemodynamics for the normal and the failing LV conditions are shown in Table I. The LVAD and the physiological controller were then switched on. For the Frank-Starling control system, an initial sensitivity of 0.47 L/min/mmHg was chosen to obtain a normal mean aortic pressure (MAP) of approximately 100mmHg. For the fixed speed control, the speed was set at 2460 revolutions per minute (RPM) to obtain the same initial haemodynamics. After 30 seconds, the system was then disturbed by a step increase in pulmonary vascular resistance (PVR) from 20 MPa.s/m^3 (Stage 1) to 60 MPa.s/m^3 (Stage 2), reducing venous return to the left atrium. At 90 seconds, the resistance was then decreased from 60 to 7 MPa.s/m^3 (Stage 3), increasing venous return to the left atrium. The changes in MAP, LAP and cardiac output (CO) were observed and compared in all five scenarios.

III. Experimental results

The results of the study are detailed in Table I. The variations in PVR firstly reduced, and then increased, the venous return to the left atrium. The Frank-Starling control system responded in a similar manner to the simulated normal LV by automatically varying the pump speed in order to obtain similar changes in MAP (92.7 to 50.9 to 113.5 mmHg) and flow rate (4.2 to 2.8 to 4.7 L/min). In contrast, there was very little change in flow rate and aortic pressure in the fixed speed pump. This constant pump flow, combined with reduced venous return during Stage 2, resulted in the approach of ventricular suction (LAP = 0) during this stage.

TABLE I

Haemodynamic values for various testing protocols in a steady state situation (1), experiencing a reduction in LA venous return (2) and an increase in LA venous return (3).

Stage	MAP (mmHg)	LAP (mmHg)	CO (L/min)	ω (RPM)
Normal LV				
1	99.0	8.6	4.8	-
2	56.0	5.2	3.4	-
3	105.0	9.2	5.1	-
Failing LV (unsupported)				
1	48.0	10.4	3.0	-
2	40.0	7.5	2.6	-
3	52.7	12.16	3.3	-
Failing LV (supported, fixed speed)				
1	94.4	6.0	4.2	2460
2	87.8	0.0	3.9	2460
3	97.2	8.8	4.3	2460
Failing LV (supported, C On, A Off)				
1	92.7	6.2	4.2	2460
2	50.9	3.5	2.8	1740
3	113.5	7.2	4.7	2700
Failing LV (supported, C On, A On)				
1	95.7	6.3	4.2	2460
2	48.2	4.1	2.6	1650
3	107.0	7.9	4.5	2580

MAP = mean aortic pressure, LAP = left atrial pressure, CO = cardiac output[a], mmHg = millimeters of mercury, L/min = litres per minute, ω = pump speed, RPM = revolutions per minutes. [a]Cardiac output during supported simulations only consisted of LVAD flow

Figure 3 shows the sensitivity of the LVAD flow rate to changes in LAP for all three methods of LVAD control. As expected, the sensitivity of the LVAD operated at constant speed was significantly smaller than that of the controlled LVAD, which resulted in ventricular suction during low venous return. In contrast, no suction or overfilling was observed with the Frank-Starling controller. The addition of the autosensitivity control varied the sensitivity from 0.47 to 0.35 to 0.41 L/min/mmHg in order to maintain the operating point within the IOZ. The use of this IOZ ensured that the operating point remained away from the point of imminent suction, whilst avoiding ventricular overfilling.

Figure 3 also shows that the control system did not track the target LVADQ perfectly, due to the use of a low-pass filter on the LVADQ.

IV. Discussion

The Frank-Starling LVAD control mechanism successfully varied pump speed in response to changes in preload, safely avoiding suction and overfilling. This implies that this mechanism can cope with the daily variations in venous return caused by postural changes and straining. The addition of the autosensitivity mechanism successfully varied the sensitivity in order to maintain the pump operation within a predefined IOZ.

In this study the IOZ was set arbitrarily in order to show the effect of the autosensitivity control and to safely operate the pump away from suction. In reality, the choice of IOZ will depend on type of LVAD therapy as determined by the clinician. A tall, thin IOZ would maintain the LAP within a small range and may be useful during initial operation and stabilisation of the device. Gradual widening of the IOZ would allow the ventricles to undergo changes in volume with changes in preload, which may be useful in ventricular recovery.

Moscato *et al.* (2010) developed a similar control strategy in which the target LVADQ was linearly related to end-diastolic LV pressure (another measure of preload), to maintain a constant afterload impedance [14]. This impedance, effectively the preload sensitivity, is adjusted by a clinician. In comparison, our control strategy incorporates an autosensitivity mechanism that ensures the operating point remains in a predefined region by automatically varying sensitivity. The only clinical input required is the initial selection of the IOZ, reducing the patient dependence on clinical staff and therefore making this control mechanism appropriate for patients discharged from hospital care.

The preload sensitivity of the control system (which varied between 0.35 and 0.47) is of the same order of magnitude as the native left ventricle, which has been reported to be between 0.213 and 0.9 L/min/mmHg [2, 3]. It is also significantly higher than that of a fixed-speed LVAD, reported as 0.105 L/min/mmHg [2]. The range of sensitivities used by the control system will depend on the clinician's choice of IOZ.

It can be seen in Figure 3 that the PID controller does not track the target LVADQ perfectly. This may pose a problem when the system is operating with higher sensitivities because the flow rate may not decrease fast enough to avoid suction during low venous return. Further investigation into the use of non-linear adaptive control techniques is being undertaken by the group to minimize this tracking error. The use of a moving average filter instead of a low-pass filter may also improve the time-domain response of the control system.

The main limitation of this control strategy is the reliance on both a pressure and a flow sensor. Implantable flow sensors consist of a flow probe clamped around an outlet cannula, and are part of the HeartAssist series of LVADs [15], however the long term reliability of these sensors is questionable [16]. There are some left atrial pressure sensors currently in development [17], but none yet commercially available. An alternative to using a direct measurement of LAP would be to use the pulsatility of the flow or pump current signals, caused by residual ventricle contractility, as an indicator of preload [10]. However, if the ventricle has no residual contractility, then alternative LAP estimation techniques will have to be developed.

V. Conclusion

In this study, a Frank-Starling-like control system for rotary LVADs is presented. This system varies the target LVADQ in response to changes in venous return. This system was evaluated in a mock circulation loop by comparing its response to an LVAD operated at fixed speed. The control strategy restored preload sensitivity to 0.47mmHg, the same order of magnitude as the native LV. The addition of an autosensitivity mechanism maintained the operating point within desired limits, improving the safety of this mechanism. The control strategy is limited by the requirement of an implanted pressure sensor in the left atrium. It may, however, be possible to use parameters such as pulsatility of pump signals as alternative control inputs.

VI. Acknowledgment

This project was supported in part by The Medical Advances Without Animals Trust (MAWA), The Prince Charles Hospital Foundation and the Critical Care Research Group.

VII. References

[1] E. Rose, A. Moskowitz, M. Packer, J. Sollano, D. Williams, A. Tierney, et al., "The REMATCH trial : Rationale, design and end points," Ann Thorac Surg, vol. 67, pp. 723-730, 1999.

[2] R.F. Salamonsen, D.G. Mason, and P.J. Ayre, "Response of Rotary Blood Pumps to Changes in Preload and Afterload at a Fixed Speed Setting Are Unphysiological When Compared With the Natural Heart," Artificial Organs, vol. 35, (no. 3), pp. E47-E53, 2011.

[3] M. Ursino, "Interaction between carotid baroregulation and the pulsating heart: a mathematical model," Am J Physiol Heart Circ Physiol, vol. 275, pp. H1733-47, 1998.

[4] R. Idema, A. van den Meiracker, A. Balk, E. Bos, M. Schalekamp, and A. Man in't Veld, "Abnormal diurnal variation of blood pressure, cardiac output, and vascular resistance in cardiac transplant recipients," Circulation Research, vol. 90, pp. 2797-2803, 1994.

[5] D.P. Veerman, B.P.M. Imholz, W. Wieling, K.H. Wesseling, and G.A. van Montfrans, "Circadian Profile of Systemic Hemodynamics," Hypertension, vol. 26, pp. 55-59, 1995.

[6] R.J. Rodeheffer, G. Gerstenblith, L.C. Becker, J.L. Fleg, M.L. Weisfeldt, and E.G. Lakatta, "Exercise cardiac output is maintained with advancing age in healthy human subjects: cardiac dilatation and increased stroke volume compensate for a diminished heart rate," Circulation Research, vol. 69, pp. 203-213, 1984.

[7] A. Tanaka, M. Yoshizawa, P. Olegario, D. Ogawa, K. Abe, T. Motomura, et al., "Detection and Avoiding Ventricular Suction of Ventricular Assist Devices," in Proc. Engineering in Medicine and Biology Society, 2005. IEEE-EMBS 2005. 27th Annual International Conference of the, 2005, pp. Pages.

[8] M. Vollkron, H. Schima, L. Huber, R. Benkowski, G. Morello, and G. Wieselthaler, "Development of a Suction Detection System for Axial Blood Pumps," Artif Organs, vol. 28, (no. 8), pp. 709-716, 2004.

[9] K. Reesink, A. Dekker, T.V.d. Nagel, C. Beghi, F. Leonardi, P. Botti, et al., "Suction Due to Left Ventricular Assist: Implications for Device Control and Management," Artif Organs, vol. 31, (no. 7), pp. 542-549, 2007.

[10] S. Choi, J.E. Antaki, R. Boston, and D. Thomas, "A sensorless approach to control of a turbodynamic left ventricular assist system," Control Systems Technology, IEEE Transactions on, vol. 9, (no. 3), pp. 473-482, 2001.

[11] A. Ferreira, J.R. Boston, and J.F. Antaki, "A Control System for Rotary Blood Pumps Based on Suction Detection," Biomedical Engineering, IEEE Transactions on, vol. 56, (no. 3), pp. 656-665, 2009.

[12] A.C. Guyton and J.E. Hall, Textbook of Medical Physiology, Philadelphia, Pennsylvania: Elseiver Inc., 2006.

[13] D.L. Timms, S.D. Gregory, N.A. Greatrex, M.J. Pearcy, J.F. Fraser, and U. Steinseifer, "A Compact Mock Circulation Loop for the In Vitro Testing of Cardiovascular Devices," Artificial Organs, vol. (Accepted: July 2010), 2011.

[14] F. Moscato, M. Arabia, F.M. Colacino, P. Naiyanetr, G.A. Danieli, and H. Schima, "Left Ventricle Afterload Impedance Control by an Axial Flow Ventricular Assist Device: A Potential Tool for Ventricular Recovery," Artif Organs, vol. 34, (no. 9), pp. 736-744, 2010.

[15] G.P. Noon, D.L. Morley, S. Irwin, S.V. Abdelsayed, R.J. Benkowski, and B.E. Lynch, "Clinical experience with the MicroMed DeBakey ventricular assist device," Ann Thorac Surg, vol. 71, (no. 90030), pp. S133-138, March 1, 2001 2001.

[16] M.S. Slaughter, C.R. Bartoli, M.A. Sobieski, G.M. Pantalos, G.A. Giridharan, R.D. Dowling, et al., "Intraoperative Evaluation of the HeartMate II Flow Estimator," The Journal of Heart and Lung Transplantation, vol. 28, (no. 1), pp. 39-43, 2009.

[17] B. Fritz, J. Cysyk, R. Newswanger, W. Weiss, and G. Rosenberg, "Development of an Inlet Pressure Sensor for Control in a Left Ventricular Assist Device," ASAIO J, vol. 56, (no. 3), pp. 180-185, 2010.