Variable Contraction Timing for a Soft Robotic Cardiac Assist Device

C.J. Payne[1,2], D. Van Story[1,2], D. Bautista-Salinas[2], M. Saeed[2], T. Thalhofer[1], P.J. del Nido[2], C.J. Walsh[1], N.V. Vasilyev[2]

[1]Wyss Institute for Biologically Inspired Engineering, Harvard University USA

[2]Department of Cardiac Surgery, Boston Children's Hospital, USA
nikolay.vasilyev@childrens.harvard.edu

INTRODUCTION

Ventricular Assist Devices (VADs) are medical implants designed to augment cardiac output for patients with heart failure. We recently proposed a soft robotic VAD (SRVAD) that adopts pneumatic soft actuators to rhythmically massage the heart left ventricle (LV) via septal bracing [1]. The performance of SRVADs is highly dependent on effective temporal synchronization between the device and the native heart contraction [1, 2]. In our previous work, we used discrete on/off valves to control airflow into the SRVAD actuators, which places inherent limitations on the dynamic response of the soft actuators. Here, we use a proportional electromechanical valve with a variable aperture for controlling the working fluid flow rate to enable arbitary actuator contraction profiles. We experimentally derived valve actuator parameters to obtain a range of actuator contraction profiles and tested these profiles *in vivo*.

MATERIALS AND METHODS

Our device and control system are thoroughly described in [1]. McKibben-based soft actuators are attached to the extracardiac bracing frame that is positioned around the LV. The frame is braced against the LV via an intracardiac bar that passes through the LV wall and is attached to the ventricular septum via a anchoring system (Fig. 1).

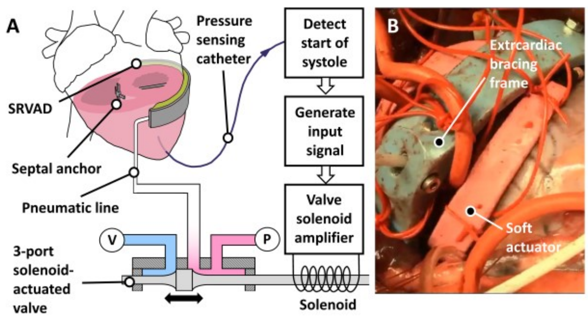

Fig. 1. (a) SRVAD, control system and (b) *in vivo* use

When cyclically pressurized, the actuators contract, which causes rhythmic loading of the LV to aid blood ejection. We adopt the same control system as previously described in [1], which uses a pressure-sensing catheter to detect the beginning of the LV systole and trigger device actuation. However, instead of using a discrete on/off valve to control valve pressurization, we adopt a 3-port proportional valve (VEF3121-2-02N, SMC) that can provide a pneumatic output of abritary forward and reverse flow rate, when connected to a regulated pressure source and a vacuum source (Fig. 1, 2). A position-controlled solenoid adjusts the valve aperture to control the output flow rate. The solenoid drive electronics (VEA250, SMC) sets the valve aperture according to an input signal (0-5V); a 5V signal corresponded to the valve being fully open to the pressure source and 0V corresponds to full vacuum. The valve is fully shut in a 2-3V deadband range, and the valve aperture is proportionally open to the intermediary voltages (Fig 2.). The McKibben-based soft actuators used in our device have a non-linear pressure-contraction relationship [2]. To account for this, our actuation strategy was to provide a high initial flow rate in to the actuator and then immediately reduce the valve aperture to retard actuator contraction. The valve input voltage (V) with respect to time (t) was governed by equation 1:

$$V = 5 - \left((5 - R_V) \left(\frac{t}{T_s} \right)^K \right) \quad \text{for } (t \leq T_s) \qquad (1)$$

Where T_s is the systolic actuation period, K is a constant which determines the rate of valve closure and R_V sets the valve aperture at the end of systole. V is then set to 0V (valve set to full vacuum) for the entirity of diastole to relax the actuator. The regulated pressure source can be adjusted to provide an input pressure, P.

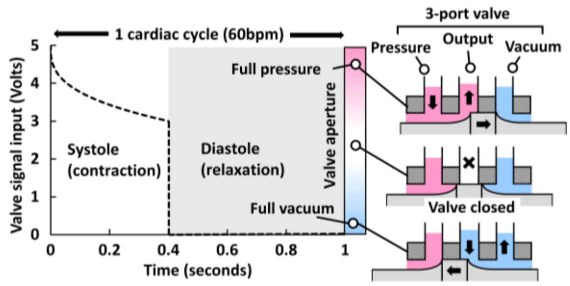

Fig. 2. 3-port proportional valve control paradigm

In a benchtop study, we emperically derived the actuation parameters (K, R_V, P) to obtain 4 distinct temporal contraction profiles within a 0.4 second time window (representing 60bpm, 40% systolic phase). Linear actuation contraction distance was measured using a potentiometer and data acquistion system, as described in [2]. We used a regulated pressure source and connected 2 McKibben actuators to the system (since two actuators are used in the SRVAD for LV support, although contraction measurements were only made on a single actuator). We used a 2m length of ¼" tubing from the valve output to the actuators. The McKibben actuators (140mm in total length) were based on a ½" mesh, with a thermoplastic bladder and nitrile oversheath to aid device recoil, as described in [2]. The actuation parameters for the actuation profiles are listed in Table 1.

We tested these 4 actuation profiles in an IACUC-approved *in vivo* porcine study (n=1) to assess the influence of actuator contraction profile on cardiac output. The device was deployed as described in [1] and the actuators were sutured to LV free wall. After SRVAD implantation, heart failure was created by injecting polystyrene microbeads of a nominal diameter of 50 to 100 µm (Megabead NIST, Polysciences Inc.) in to the coronary arteries to induce LV ischemia. During heart failure, the device was actuated for each contraction profile and the aortic blood flow was measured using a flow probe (20PS, Transonics Corp). The control system regulated T_s to be 40% of the cardiac cycle. Data analysis was based on n=7 consecutive cardiac cycles.

RESULTS

Temporal actuator contraction profiles and contraction velocity profiles are presented in Fig 3. Key hemodynamic data is presented in Table 1. The animal heart rate was 61-67bpm during actuation periods. In vivo aortic flow data is presented in Fig 4.

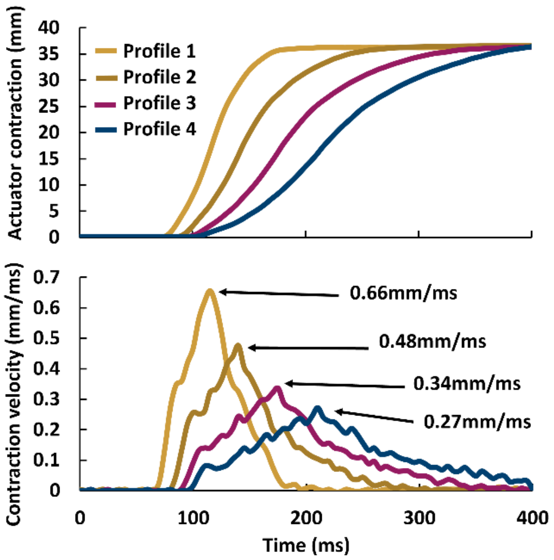

Fig 3. 140mm McKibben soft actuator contraction profiles and corresponding contraction rates (profiles 1-4).

Profile	K	Rv	P (psi)	Mean flow rate (L/min)	Peak LV pressure (mmHg)
1	0.15	3.0	21	0.97	46.8
2	0.01	5.0	15	1.06	43.7
3	0.01	3.9	15	1.03	47.3
4	0.005	3.62	17	1.25	52.0

Table. 1. Actuation parameters and hemodynamic data for each contraction profile tested

DISCUSSION

Fluidic-based soft robotic systems commonly use on/off valve switching and static-pressure based control paradigms for device actuation. Such an approach is sufficient for many soft robotic applications, but in cardiac assist applications the dynamic response of the actuator is critical to device efficacy. The proposed control methodology has demonstrated the ability to tune the actuator response time to aid SRVAD-heart synchronization.

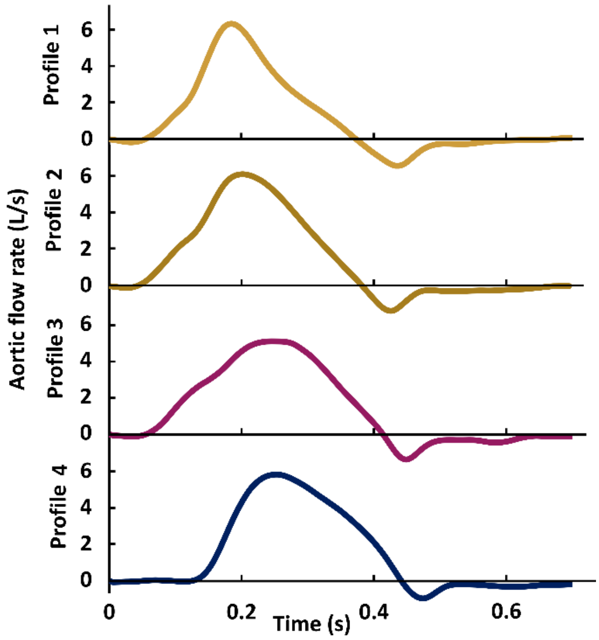

Fig 4. Systolic phase of aortic flow traces during actuation of the SRVAD for the 4 actuator contraction profiles (average aortic flow profile is based on n=7 consecutive cardiac cycles).

Bench testing of the control system demonstrated the ability to achieve a broad range of contraction profiles. There was a 2-fold increase in full contraction response time between the extreme profiles (1 and 4) and a ×2.4 reduction in peak contraction velocity. *In vivo* testing demonstrated that the different actuator contraction profiles distinctly alter the aortic blood flow rate trace (Fig. 4). Nonetheless, cardiac output was comparable between actuator contraction profiles. Profile 4 (the slowest actuation profile) facilitated greater cardiac output and higher peak LV pressures. This agrees with prior work with direct cardiac compression devices that demonstrated that slower actuator contraction is preferable to faster contraction in maximizing cardiac output [2], but further investigation is needed to draw definitive conclusions. Future work will consider modeling of the soft actuator dynamic system to allow prediction of the optimal actuation parameters. This work will need to consider the transfer function of the electromechanical valve and fluid flow conditions. It will also be desirable to model the dynamic response of the actuator when coupled to the heart.

REFERENCES

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