EFFECT OF CONTINUOUS FLOW BIVENTRICULAR ASSIST DEVICE ON PRESSURE-VOLUME LOOP: A SIMULATION STUDY

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ABSTRACT
Biventricular assist device (BiVAD) support is a mechanical circulatory support using both left ventricular assist device (LVAD) and right ventricular assist device (RVAD). BiVAD support is now increasingly used as an alternative treatment for end-stage heart failure patients, such as bridge to transplantation, bridge to recovery, and permanent support. Both flow balancing and Starling response of the right and left ventricles should be concerned to prevent the suction event on both right and left ventricles. In this study, the cardiovascular system model using the rotary blood pump was implemented to study the mechanical circulatory support management during left ventricular assist device (LVAD) support alone and during BiVAD support. The pathological ventricles were generated by reducing the maximum elastance (Emax) in the model. The pressure-volume loop of difference heart conditions (normal ventricle: 100% of Emax, pathological ventricle: 50% of Emax) and the levels of pump support (non-support, partial support and full support) were simulated. From all simulations, the suction events in the right ventricle (represented by negative values of right ventricular volume and pressure) were observed during the unbalanced flow conditions of BiVAD support. Therefore, the flow balancing or pump speed adjustment is a key for BiVAD management.

KEY WORDS
Biventricular assist device, BiVAD, Rotary blood pump, RBP, BiVAD management

1. Introduction
Left ventricular assist device (LVAD) has been accepted as an alternative treatment for end-stage heart failure patients [1-2]. However, after LVAD implantation, 30% of LVAD recipients clinically demonstrate a right ventricular failure and potentially need right ventricular assist device (RVAD) [3]. For both LVAD and RVAD support, it is called biventricular assist device (BiVAD). In the past, pulsatile flow ventricular assist device system was used for BiVAD application because of the volume control advantage. Nowadays, the rotary blood pump (RBP) or continuous flow ventricular assist device provides increased device reliability, smaller size, and significant improvement of pump management. Thus, RBP is selected for the second generation of LVAD. Many RBP systems were used in animal experiments for BiVAD support [6]. For clinical treatment, HeartWare HVAD (HeartWare Inc, Framingham, MA) has been successfully implanted as a BiVAD in Germany [7-8]. However, since RBP is pressure dependent, an experienced person is required for balancing the flow between right side and left side of the heart [3-5] and the training using an animal model is very costly. Therefore, the simulation that can imitate the hemodynamic during BiVAD support could be a useful tool to educate clinical staff. Additionally, each RBP from different companies has different speed-characteristic. Hence, the simulation with a correct model of each RBP could be a useful tool for training to prevent the suction event (over unloading ventricle or unbalanced flow) and to test new purpose of clinical treatment [9-12]. In this present study, the numerical simulation of heart failure patients during BiVAD support was developed.

2. Materials and Methods
In this simulation, the cardiovascular system model using rotary blood pump for both LVAD and RVAD was based on the lump parameter model such as resistance, capacitance, and inductance in electrical analog circuit. The cardiovascular system model could be divided into 4 parts as follow: 1) heart part: right atrium, tricuspid valve, right ventricle, pulmonary valve, left atrium, mitral valve, left ventricle, and aortic valve; 2) pulmonary part: pulmonary arteries, pulmonary capillaries, and pulmonary veins; 3) myocardial perfusion: coronary arteries, coronary capillaries, and coronary veins; 4) systemic vessels: systemic veins, vena cava, ascending aorta, descending aorta, and systemic arteries. The mathematical models were based on the study by Sun Y. et al [13]. The time varying elastance curves of left atrium, left ventricle, right atrium, and right ventricle were slightly modified from Liang F and Liu H [14]. To generate the level of heart failure, the cardiac contractility of both right ventricle (RV) and left ventricle (LV) were modified by the scaling factors in time-varying elastance equation of Liang F and Liu H [13]. Scaling factors reflecting the $E_{max}$ parameter in the time-varying elastance curve were designed for the nervous reflex control of both RV and LV contractility.

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The RBP model was implemented based on the experimental model of MicroMed-DeBakey LVAD (MicroMed Cardiovascular Inc. Houston, TX) of Schima group [5, 15]. The structure of RBP model was developed based on the normal construction of RBP by connecting the inflow cannula of RBP model to LV model and connecting the outflow cannula of RBP model to ascending aorta model.

The numerical model of the whole cardiovascular system including the RBP at both right and left sides of the heart were implemented in MATLAB Simulink® (MathWorks Inc, Nick, MA) [13-14]. The simulation included the normal heart (maximum elastance: 100% \( E_{max} \)), left ventricular failure without LVAD support (low contractility: 50% \( E_{max} \)), left ventricular failure with partial and full LVAD supports, right and left ventricular failure without BiVAD support, right and left ventricular failure with partial and full BiVAD supports. The parameters of systemic vascular system, pulmonary vascular system, coronary vascular system and heart valve were kept constant as a normal condition adopted from Sun Y. et al [13]. The four time varying elastance curves of the heart were taken from Ling F and Liu H [14].

The level of support was regulated by adjusting pump speed to 7.5, 9, 10, and 11 krpm (from partial support to full support) at pathological condition. Partial support means the aortic valve opens during systole. Full support means the aortic valve always closes during cardiac contraction. (7.5 krpm is the lowest speed of MicroMed-DeBakey LVAD model.) The heart conditions and pump speeds are shown in Table 1.

In this study, the pressure volume loop (PV-loop) which is a plot between pressure signal and volume signal was plotted. The PV-loop of both LV and RV in normal heart, LV failure condition alone, LV-RV failure condition, LVAD support and BiVAD support are considered.

### Table 1. Heart conditions and pump speeds

<table>
<thead>
<tr>
<th>No.</th>
<th>RV</th>
<th>LV</th>
<th>RVAD (krpm)</th>
<th>LVAD (krpm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.</td>
<td>Normal</td>
<td>Normal</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>2.</td>
<td>Failure</td>
<td>Failure</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>3.</td>
<td>Normal</td>
<td>Failure</td>
<td>-</td>
<td>7.5</td>
</tr>
<tr>
<td>4.</td>
<td>Normal</td>
<td>Failure</td>
<td>-</td>
<td>9</td>
</tr>
<tr>
<td>5.</td>
<td>Failure</td>
<td>Failure</td>
<td>-</td>
<td>7.5</td>
</tr>
<tr>
<td>6.</td>
<td>Failure</td>
<td>Failure</td>
<td>-</td>
<td>9</td>
</tr>
<tr>
<td>7.</td>
<td>Failure</td>
<td>Failure</td>
<td>7.5</td>
<td>7.5</td>
</tr>
<tr>
<td>8.</td>
<td>Failure</td>
<td>Failure</td>
<td>9</td>
<td>9</td>
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<tr>
<td>9.</td>
<td>Failure</td>
<td>Failure</td>
<td>7.5</td>
<td>9</td>
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<tr>
<td>10.</td>
<td>Failure</td>
<td>Failure</td>
<td>7.5</td>
<td>10</td>
</tr>
<tr>
<td>11.</td>
<td>Failure</td>
<td>Failure</td>
<td>7.5</td>
<td>11</td>
</tr>
</tbody>
</table>

Normal: 100% of \( E_{max} \), Failure: 50% of \( E_{max} \), RV: right ventricle, LV: left ventricle, RVAD: right ventricular assist device, and LVAD: left ventricular assist device.

### 3. Results

The hemodynamic simulations of both normal LV and normal RV were shown in Figure 1 (RVP: right ventricular pressure, PAP: pulmonary artery pressure, LVP: left ventricular pressure, AoP: aortic pressure). The normal and pathological LV conditions were confirmed by the pressure volume loops during LVAD support at the difference levels of \( E_{max} \) (100% and 50% of normal condition) and shown in Figure 2. The pressure volume loops of normal RV and pathological LV with LVAD support at 7.5 and 9 krpm (partial and full support) were shown in Figure 3. In normal RV, the effect of LVAD support was not significant. During BiVAD support, the pressure volume loops of pathological heart at the same constant speed of VAD (RVAD and LVAD at 7.5krpm) were shown in Figure 4. In Figure 5, the pressure volume loops of higher pump speed were shown (RVAD and LVAD at 9 krpm). The suction event was observed in this condition. The pressure volume loops of varied speeds of LVAD were shown in Figure 6.

### 4. Discussion

Based on an animal experiment with LVAD support [16] and clinical reports with LVAD support [17-18], the hemodynamic changes mainly depend on the level of LVAD support (partial and full supports) and the level of heart failure (low cardiac contractility or low \( E_{max} \)) which are similarly demonstrated in this study in Figure 1-3. In Figure 1, the drift of right ventricular pressure arose from the effect of breathing cycle.
In Figure 2, the reduction of the slope between pressure and volume is similar to the data from the animal experiment [16]. The reduction of area inside pressure volume loops mainly depends on the increasing level of LVAD support that also corresponds to the animal experiment [16].

In Figure 3, the pressure volume loops of normal RV mainly showed the small reduction of RV volume from partial LVAD support to full LVAD support. Therefore, LVAD did not significantly affect normal RV. For both-sided heart failure, BiVAD support showed significant changes of ventricular volumes from non-support to partial support at 7.5 krpm. The RV volume reduced and LV volume increased in Figure 4. The unloading RV by pump increased the blood flow which increased the left ventricular end diastolic volume (LV-EDV) from 100 ml to 200 ml. In Figure 5, the unbalance of flow showed the suction event at 9 krpm of pump speed support on both RVAD and LVAD. In Figure 6, the flow management was implemented by using the lowest speed of RVAD and varied speeds of LVAD (RVAD speed: 7.5krpm and LVAD speed: 7.5, 9, 10, and 11 krpm). The pressure
volume loops of LV reduced depending on the LVAD speed and the RV volume increased following the change of LVAD speed. The increase of right ventricular end-diastolic volume depends on the venous return related to the level of LVAD support. However, the lowest speed of RVAD support still increased the LV-EDV from 100 ml to 200 ml which is the abnormal support condition. Additionally, the increase of RVAD speed is also possible to generate the suction event. Therefore, the normal pump that is designed from LVAD needs a BiVAD management for flow balancing. The original RBP that is designed for LVAD support still increased the LV-EDV from 100 ml to 200 ml which is the abnormal support condition. However, the lowest speed of RVAD support may not be suitable for RVAD purpose. The LV-RBP modification is needed for RVAD application [3].

5. Conclusion

The computer simulation of cardiovascular system and BiVAD system suggests that the flow balancing of BiVAD is a key of BiVAD management.

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References