Computer Model of An Electro-hydraulic Left Ventricular Assist Device

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INTRODUCTION

A left ventricular assist device(LVAD) is defined as a mechanism which operates parallel with the natural heart. Mainly, the objects of LVAD for acute or temporary use are to maintain systemic arterial perfusion, to reduce heart workload and to improve cardiac oxygen balance for recovery of a failing heart. There are many different LVAD configurations and operating modes. (eg. Ventriculo-Aortic (V-A), Atrio-Aortic (A-A) bypass and pulsatile-synchronous, asyncronous, non pulsatile flow etc)

Many good cardiovascular system models were developed differently [1].[2]. For intraaortic balloon pump(IABP)[3], several researchers modeled and indicate that the timing of the inflation and deflation of the balloon is critical[4].[5]. In pneumatic LVAD models [6].[7].[8], LVADs were simplified as one pressure source and one capacitor without inertance component for cannular, volume limit of the blood sac and micro-controller.

In this study we modeled an electro-hydraulic LVAD which could synchronize with natural heart phase (R peak of ECG) and control the assist blood volume. The LVAD model was incorporated with circulatory model to study the interaction between the device and the natural heart especially concerning the cardiac output, oxygen balance of the injured myocardium under various control modes of LVAD . Using this model, the effect of the A-A bypass and the V-A bypass were compared. This model is used as a helpful tool to test the new control schemes or other parameters before animal experiments. Energy efficiency was considered in each component.

METHODS

The electro-hydraulic LVAD that we developed before is shown as a schematic diagram in **Fig**.1. Its mechanism is that motor rotation is converted into translation of the pusher plate connected to the pumping sac by a ball screw. The hydraulic energy generated by the motor makes the deflation of the blood sac,after which the pump ejection starts. Our LVAD could operate with a relatively exact delaytime than conventional pneumatic LVADs. The computer model consists of three parts : a) a cardiovascular model ,b) a PI-control model of the motor and c)a model of the hydraulic parts of LVAD.

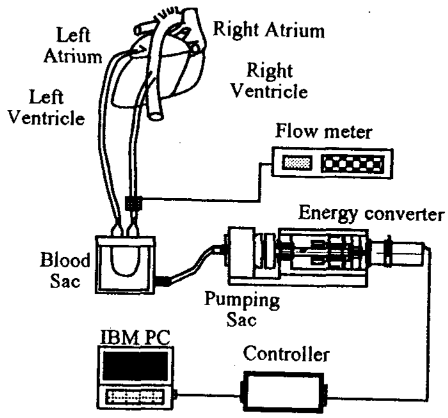

Fig.1 Electro-hydraulic Left Ventricular Assist Devce

Cardiovascular System Model

The cardiovascular model similar to that of Rideout[9] is shown in **Fig**.2. Originally, it is a Windkessel model, i.e. a network of a resistance, an inductance and a compliance. It has a baroreceptor feedback model which controls heart rate and elastance of the heart. The principle of the baroreceptor model is that the gain is determined by the mean aortic pressure. Small modifications were made to evaluate the hemodynamic and metabolic effects of heart.The oxygen availability (Oa) is represented by the time integral of coronary flow per heart beat. The stenosis of coronary vessel was characterized by setting resistance to 10 mmHg*sec/ml.[10].[11]. The heart workload was assumed proportional to the pressure volume area(PVA) and maximum elastance of left ventricle(MEV). It represents the oxygen consumption of heart.[12]. Relative comparisons of each value were made while varying the LVAD control parameters.

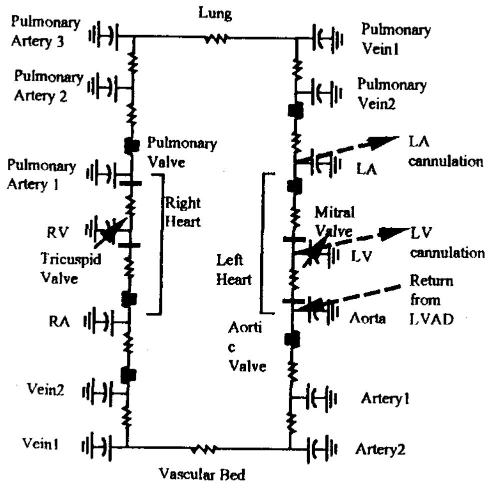

Fig.2 Cariovascular Model

LVAD System Model

In **Fig**.3, the LVAD system is shown by electrical components, where the motor and pumping sac are simplified to a variable flow source, which is different from other computer models.

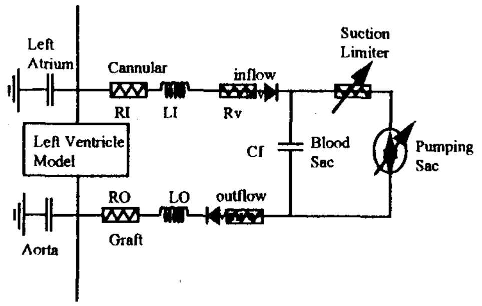

Fig.3 Hydraulic System Model of LVAD

-Motor Control System

PI control is obtained by a micro-controller (i80196). From the encoder signal,the velocity of motor was controlled by changing the voltage with PWM(Pulse Width Modulation) technique. Motor equations follow as below. Each millisecond, the control values are updated.

$i = (e-Ke*w)/R$

$dw/dt = (Kt*i-Pb*G)/J$

i : current [A] e : voltage [V]

Ke : back emf constant [V/rad*sec] w : motor velocity [rad/sec]

Kt : torque contant [kg-cm/A]

Pb : pumping sac pressure [mmHg]

G : conversion factor J : rotational inertia [g-cm-sec2]

$e= A*err + B*\text{sum of err}$

err= ref- encoder signal(1000 pulse/rev)

A : proportional gain B:integral gain

ref : parabolic velocity profile with respect to the stroke length

-Hydraulic Components Model

Blood sac, chamber, inlet outlet cannular are similar to the conventional pneumatic LVAD, except that the working media is not air but fluid. The cannular is modelled by an inductance and a resistance.The values are calculated considering turbulent flow and tapering shape[13].[14]. The artificial valve is treated as an ideal diode and a pressure varying orifice resistance which makes the pressure drop proportional to the square of flow.[15]. The pumping sac compliance and the effective area were determined experimentally. We implemented Suction- Limiter which is composed of a collapsible tube with a static negative pressure outside. If the LAP is too low,the tube will be deflated. Therefor, the amplification of current can be detected by the controller. the Suction-Limiter is modeled as a pressure varying resistance. The values of each component are shown in **Table** 1

Table 1 Paramer Values of LVAD

RI	0.262	RO	0.082	(mmHg/ml*sec)
LI	0.160	LO	0.142	(mmHg/ml*sec^2)
Rv	0.0002252			(mmHg/ml^2*sec^2)
Cf	0.0532-0.003988*(1000+Position)			(ml/mmHg)

Position =1000* # of motor rev.

Numerical Solving Method

In total, 44 differential equations are solved by 4th order Runge-Kutta method with 1 ms time interval. Using a monitor program, the pressure and flow waveform can be displayed graphically. The validity of the model was tested by comparing the waveforms of pressure, voltage and current from animal experiment with data generated by computer model.

Simulation

Pulsatile perfusion may improve myocardial perfusion and metabolism. [16]. In deep anesthesia, the baroreceptor feed back may be decreased. Synchronized A-A bypass and V-A bypass were investigated with feedback and without while varying the delaytime of VAD ejection.

RESULT

Fig.4 shows waveforms of the aortic pressure and VAD assist flow during the animal experiment. A hardcopy of the monitor program is shown in **Fig**.5. In A-A bypass, the ejection just after R peak increases the oxygen availability and decreases the oxygen consumption in **Fig**.6.The result agrees with other reports in conventional pneumatic VAD with triangular shape ejection pressure.[7]. The feedback simulation result is not so different from the simulation with no feedback but both cardiac output and pump output were decreased because of the heart rate regulating to 75 bpm, see **Fig**.7. The V-A bypass results show dramatic changes as a function of the delay time. Around 0.3 sec delaytime, the cardiac output was reduced because the natural heart could not compete with the LVAD in **Fig**.8. A certain shift in the delaytime axis is shown, see **Fig**.9. The energy transmitted to the cardiovascular system was calculated as shown **Fig**.10. The energy losses may be generated by the pulsatility of flow and pressure,which are transferred into heat.

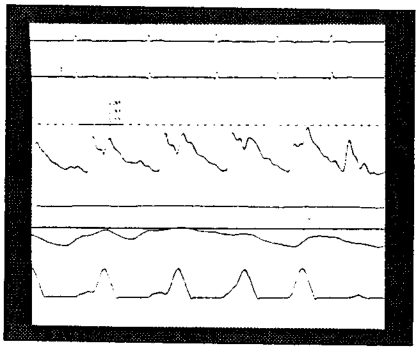

Fig.4 The Waveform of Pressure of Animal Experiment

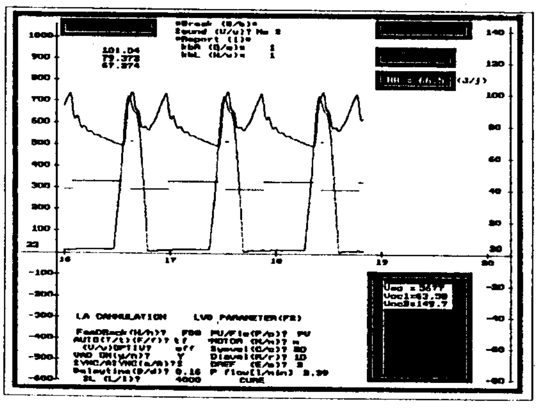

Fig.5 Monitor program

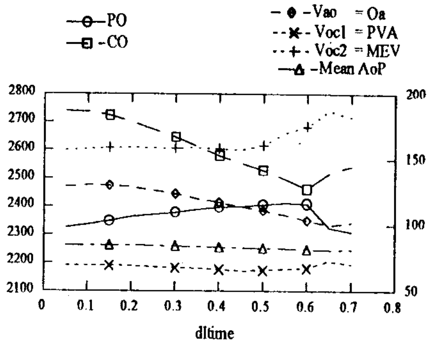

Fig.6 A-A bypass without feedback

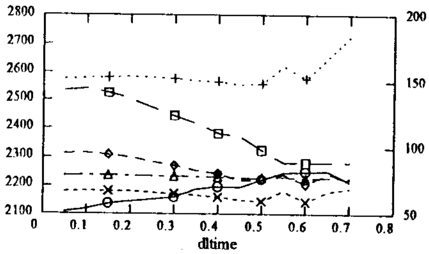

Fig.7 A-A bypass with feedback

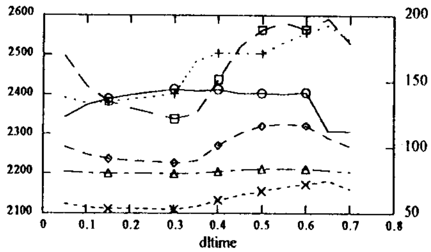

Fig.8 V-A bapass without feedback

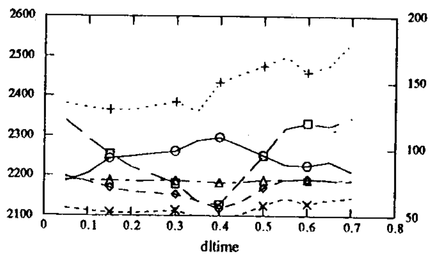

Fig.9 V-A bypass with feedback

DISCUSSION

In the A-A bypass to increase oxygen availability, the electro-pump must eject just after R peak signal, i.e. that maximum ejection may occur after aortic valve closure, which can make a relatively high aortic pressure causing the coronary flow to increase. The same concept was applied IABP inflation time. With respect to oxygen consumption, it is important not to increase the afterload of heart and to reduce the ventricular pressure. The pump ejection at the end of the diastolic phase of the heart must be avoided, which increases aortic pressure at systolic start. As the preload of the left ventricle is dependent on pump suction time, the pump uptake time must be adjusted to maintain the left atrial pressure within the physiological range ,especially to prevent the atrial from collapsing. In addition, Long term use of LVAD may develop arterial hypertension due to low left atrial pressure.[17].

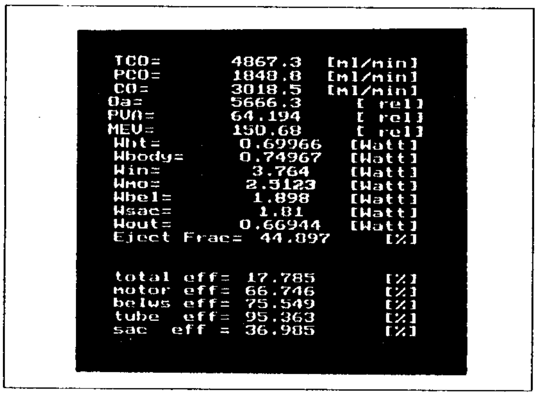

Fig.10 Energy Consideration

The V-A bypass causes a large decrease in PVA, since the direct uptake from ventricle reduces the stroke volume. Besides this it can offer better inflow to the pump than the A-A bypass, which may result in more assist flow. But the physical restriction such as damage to the myocardium may limit the use of the V-A bypass. If the failing heart must recover, the general health of the ventricle is more important than the active unloading effect and a large pump output. Too much unloading by complete decompression of the ventricle may result in excessive stress of the myocardium.[18].

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