Computer Simulation of Electrohydraulic Left Ventricular Assist Device

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Introduction

Approximately 1.0 % of open heart surgery patients become unweanable from cardiac bypass during the surgical procedure. In addition nearly 20 % of patients accepted for cardiac transplantation die while waiting for a donor heart. Ventricle assist devices (VAD) provides one of the realistic solution to these dilemmas. [1]

A newly designed electrohydraulic VAD was developed under two design philosophies: (1) portability (2) controllability. Though sufficient assistance volume of blood observed during animal experiments, the suction pressure surges was detected occasionally in the blood sac which caused the severe blood trauma. Long cannula was assumed to be one of the major causes of the pressure fluctuations due to increased inertia of the fluid flow. To regulate the peak pressure, some control parameters and design variables must be changed.

We developed a mathematical model of the fluid conduit of VAD and a simple mock circulatory system (MCS) to simulate the operation conditions and optimize the design variations in the point of view to prevent pressure surges. Throughout the computer simulation, we could find some possible solution to improve the performance of the present VAD.

Materials and Methods

System Description of VAD

A simple MCS which is composed of an aortic chamber, an inlet chamber with a flow meter is shown in Fig. 1. The Electrohydraulic VAD composed a blood sac, an energy converter system, and the controller with IBM PC is also shown in Fig. 1.

Computer Model

The mathematical model was built by using four conventional Windkessels. The energy converter consisted in a bellows, a ball screw and an electric motor was modelled by a fluid flow source. Two Windkessels coupled in a closed system with the blood sac represents the MCS. The schematic diagram of the model is shown in Fig. 2. A set of state-variable equations was obtained from the above model. The state equation was composed of eight state variables in a nonlinear first order differential type which included heart valve function and sac deformation characteristics. The nonlinear differential equation was solved numerically by using the fourth order Runge-Kutta method with calculating time step of 0.03 msec. Simulation was performed in C language on IBM 486 PC.

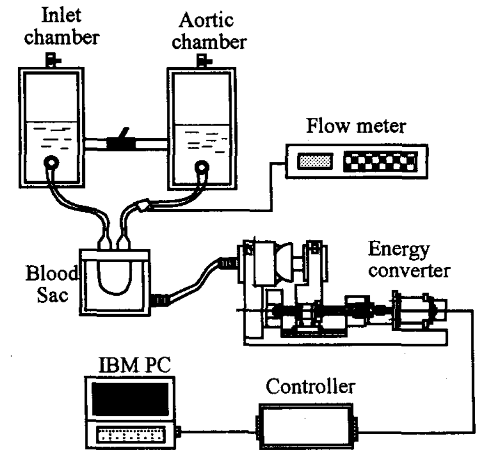

Fig 1. LVAD Mock System

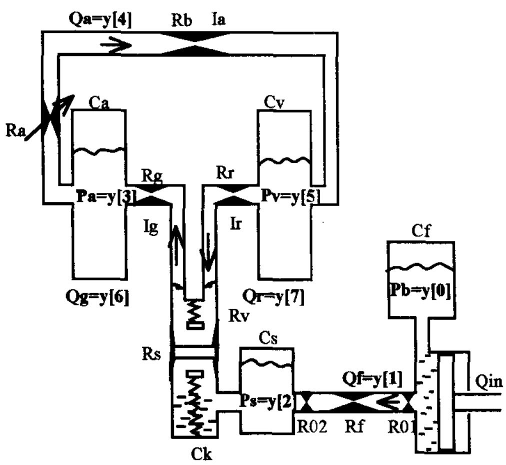

Fig2. Model of LVADMock System

Results and Discussion

Fig. 3 shows the relationship between the cardiac output (C.O.) and the inlet pressure when aortic pressure (AoP) is 120mmHg. Fig. 4 shows the relationship between C.O. and AoP when the inlet pressure is 10mmHg. Fig. 5 shows the relationship between C.O. and the stroke length of energy converter. From these data, we insist that the assistant C. O. of our VAD can be controlled by the stroke length independent of the change of pre-load and after-load. The change of C.O. is proportional to the change of the stroke length which is one of the control parameters.

The simulated results was similar to the data measured in the MCS test. The dimension of components will be modified to optimize the performance of the VAD and to prevent highly negative suction pressure generation.

In order to simulate more realistic valve function and sac deformation characteristics in the model, we must change the state variables and introduce other nonlinear functions in addition to other solving methods for the nonlinear state-variable equations.

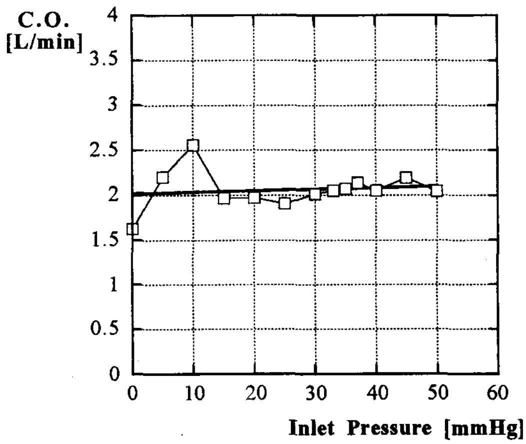

Fig3. Effect of Inlet Pressure to Simulated Cardiac Output of LVAD

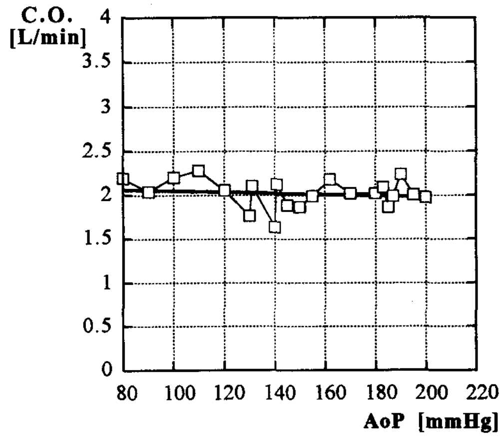

Fig4. Effect of AoP to Simulated Cardiac Output of LVAD

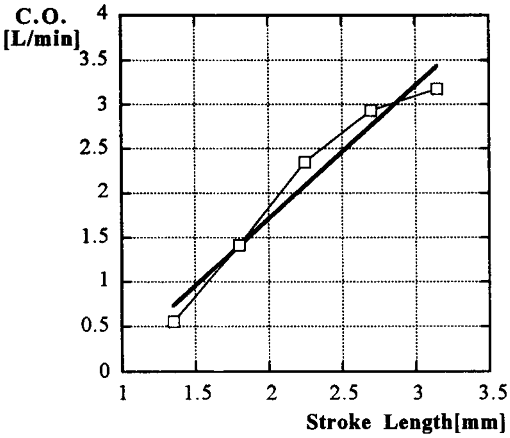

Fig5. Effect of Stroke Length to Simulated Cardiac Output of LVAD

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