

Multi-objective Optimal Control of a Heart Assist Device

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ABSTRACT

Heart assist devices play an important role in the treatment of heart disease. One of the key issues with these devices is the control of pump speed to satisfy multiple clinical requirements such as providing adequate cardiac output at appropriate left atrial pressure and systolic arterial pressure. A controller based on a multi-objective performance criterion is being developed at the University of Pittsburgh to fulfill these requirements. This paper describes the criterion and the penalty functions from which it is derived. The algorithm was applied to simulated cardiovascular data and to data from animals with implanted assist devices, and some preliminary results on the shape of the non-inferior set defined by the penalty functions were obtained.

1. INTRODUCTION

Heart disease is a major health problem in the United States and is the leading cause of death for Americans. Heart transplantation has become accepted to treat severe cases of the disease and the demand for heart transplants exceeds the supply. Mechanical ventricular assist device is often the only treatment available for patients waiting for a transplant. The goal of the assist device is to provide sufficient cardiac output at a pressure to maintain adequate perfusion of the patient's body. Initial efforts to develop assist systems have focused on critical problems related to device efficacy and safety. The control of existing ventricular assist devices (VADs) depends on human interventions. This open-loop approach may be effective in a monitored environment; however, continuous engineering and clinical support are required, and the patient's activities may be limited. This control approach may not supply adequate blood flow for ambulatory patients. As patients are supported for longer period of time, advanced controllers are needed to satisfy these requirements.

A new generation of devices being evaluated is based on turbo-hydrodynamic methods for pumping. These devices have several advantages over current devices. They are compact, efficient, simple to implant, inexpensive and reliable [1]. However they pose a more difficult control problem because the output of these pumps is sensitive to systemic input impedance. There are a limited number of such pumps used clinically, all of which are currently operated in an open-loop or "human in the loop" fashion.

We are developing an intelligent control approach for turbine assist devices to meet the clinical requirements

provided by heart transplant clinicians in the Artificial Heart Program at the University of Pittsburgh. The control structure consists of several control algorithms which include a default algorithm, two heuristic algorithms based on the assist device operating characteristics, and an optimal performance algorithm that will use an adaptive patient model and a multi-objective performance criterion. An intelligent supervisor determines the appropriate algorithm to use based on the required cardiac output, the validity of the information available, and the operating status of the device. This paper deals with use of multi-objective optimal control to define the optimal performance criterion.

2. METHODS

According to clinicians, the heart assist device control system should satisfy the following three requirements:

- 1). Cardiac output, denoted by K , should be above the minimum value required to support the activity level of the patient;
- 2). Left atrial pressure, denoted by A , should be maintained below approximately 10-15mmHg to avoid pulmonary edema and above approximately 0 mmHg to avoid ventricular suction;
- 3). Systolic arterial pressure, denoted by F , should be maintained between patient-specific limits.

We can formulate the multi-objective optimization problem using the concept of penalty function based on these clinical rules. The penalty is a function of the controlled outputs: cardiac output K , arterial pressure A , and left atrial pressure F . In general it is impossible to minimize all three penalty functions simultaneously and we must treat this as a vector-valued minimization problem [2]. In order to help clinicians to determine which pump speed to choose, we seek the non-inferior set of the pump speeds from which to select an operating speed. We define $J_v(w) = [J_1(w), J_2(w), J_3(w)]$ as a vector-valued performance measure, where w is the pump speed and $J_1(w), J_2(w), J_3(w)$ are the penalties associated with K, A and F . The non-inferior set S is defined as the set for which there exists no speed not in S which makes $J_1(w), J_2(w), J_3(w)$ smaller than values obtained using speeds in S . To find the non-inferior set, we minimize $J = \sum_{i=1}^3 \alpha_i J_i$, where $\alpha_i \geq 0$ and $\sum_{i=1}^3 \alpha_i = 1$. The detailed algorithm has been described elsewhere [2][3].

For the preliminary study described in this paper, we consider only two clinical requirements: systolic pressure ($A(w)$) and left atrial pressure ($F(w)$). Fig. 1 shows

penalty functions for them. These functions were obtained by fitting a fourth order polynomial to five points specified by clinicians. The functions specify the penalty associated with each value of systolic pressure and left atrial pressure.

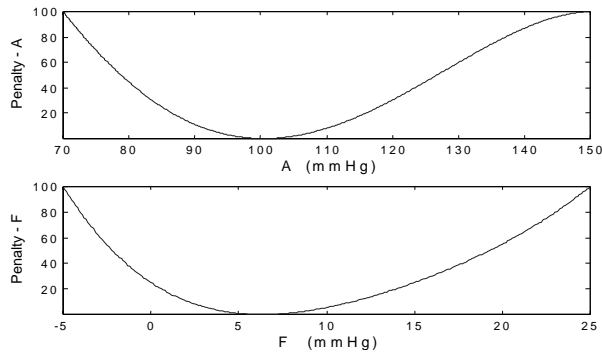


Figure 1. Penalty function for systolic arterial pressure (A) and left atrial pressure (F)

3. RESULTS

To examine the non-inferior set of the pump speeds, simulated cardiovascular data were obtained using the Physbe cardiovascular simulation program [4] and a model of an axial flow pump [5]. The simulations included ramp and step speed changes, with steps of 500 rpm. By simulating systolic pressure and left atrial pressure as functions of speed, we can calculate the functions $A(w)$ and $F(w)$. The penalty functions defined in Fig. 2 provide the relationship between A , F and the corresponding penalties. If we define the penalty functions of AOP and LAP (with respect to speed) as $J_{aop}(w)$ and $J_{lap}(w)$, then the total penalty is

$$J(w, \alpha) = \alpha J_{aop}(w) + (1 - \alpha) J_{lap}(w)$$

where α varies from 0 to 1 to reflect the relative importance of each objective.

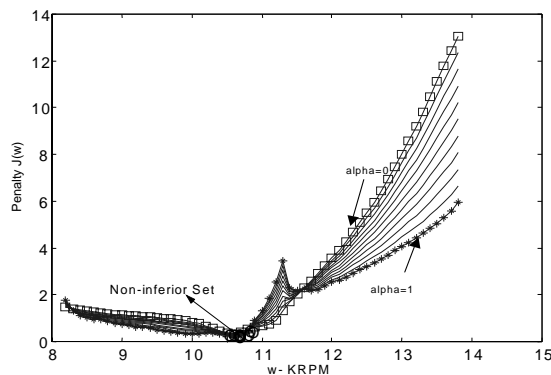


Figure 2. Combined Penalty of A and F with respect to speed based on simulation

The non-inferior set is then obtained by finding the minimum value of the combined penalty $J(w)$ for different values of α from 0 to 1. The circles in Fig. 2 show the speeds corresponding to the minimum combined penalty for different values of α . The non-inferior set was found to be between 10.5 to 10.8 KRPM. That is, as α changes

from 0 to 1, the optimal speed for $J(w, \alpha)$ varies between 10.5 and 10.8 KRPM. Whatever weighting of the two objectives a clinician desires to use, he or she should always select a speed from this range.

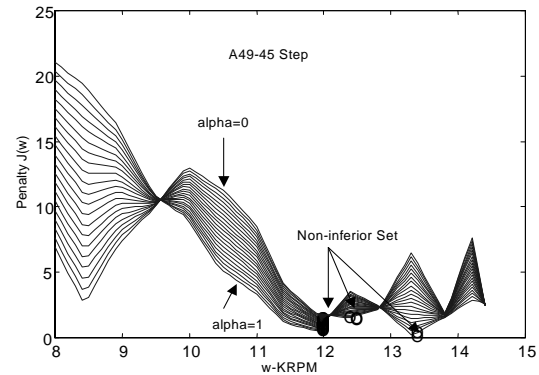


Figure 3. Combined Penalty of A and F with respect to speed based on xperiment

Animal data using a Heartmate-II (Nimbus Medical, Rancho Cordova, CA) were also available to test the approach used with the algorithm to obtain the non-inferior set for operating speeds. Fig. 3 shows the result for one animal data set. The speed was increased and decreased in steps, as was done in the simulation. The non-inferior set obtained here contains three disjoint speeds (around 12, 12.5, and 13.45 KRPM).

4. DISCUSSION

In this paper, a multiple-objective optimal control algorithm is described. The non-inferior set of pump speeds obtained here can give clinicians guidance to choose desired speed to achieve appropriate cardiac output and pressure. Also we noted in the non-inferior set obtained in Fig. 3 is disjoint. This could be caused by the noise data and non-convexity of the penalty. We will address these issues in the future.

5. ACKNOWLEDGMENT

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6. REFERENCE

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