Optimisation Based Identification Of Cardiovascular System Model Parameters For Patient Specific Diagnostic Assistance

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Introduction

There are many examples of closed loop lumped parameter cardiovascular system (CVS) models in the literature generally designed to simulate certain types of CVS function to aid understanding and diagnosis. However, methods to make these models directly useful to medical staff have not been investigated in much detail. For diagnostic purposes, a model must simulate a given person's CVS function as accurately as possible using as much patient specific information as available. Once a patient specific model is identified it can be used to assist medical staff in diagnosis and treatment selection.

Closed loop CVS models found in the literature include those of Santamore (1991) as well as Beyar et.al. (1987) that focus primarily on modelling ventricular interaction. Ursino (1999) investigates modelling carotid baroregulation, and Olansen et.al. (2000) contains an extensive model that mentions the use of parameter estimation techniques to estimate the value of key parameters.

This research employs mathematical optimisation to identify model parameters required to produce target performance characteristics. Where a patient specific model is required, the target performance characteristics will be parameters measured from a patient, such as heart rate (HR), blood pressure (BP), ventricle chamber pressures and CVS flow rates (Q).

Method

A six chamber closed loop CVS model, developed by Smith et.al. (2003), is used in all simulations. Figure 1 shows a diagram of the CVS model where resistors (R) simulate resistance effect on the flow, inductors (L) simulate inertial effects on flow around valves, and elastances (E) define the elastic properties of chambers. Inductors are not

added in the systemic (sys) and pulmonary (pul) circulations, as there is minimal change in flow rate in peripheral arteries. Lines between the left (lv) and right (rv) ventricles indicate ventricular interaction, which includes the effects of the active septum and the passive pericardium. Elastic chambers labelled vc and ao represent the vena-cava and the aorta respectively and approximate the systemic circulation system, while the pulmonary system chambers labelled pa and pu describe the pulmonary artery and the pulmonary vein.
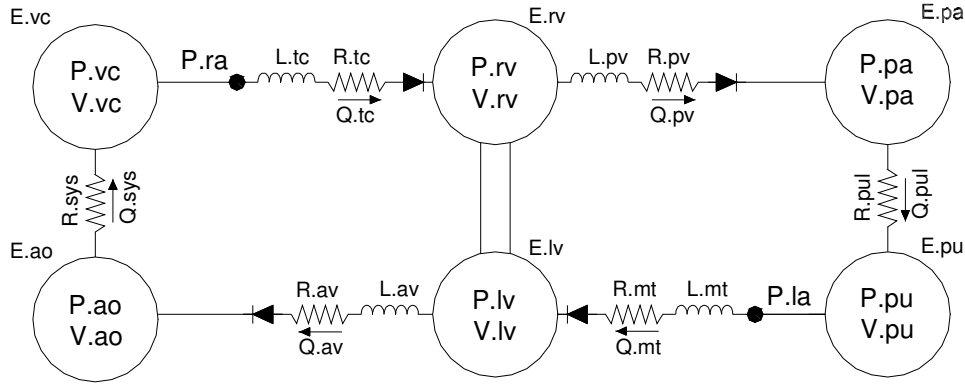

Figure 1, Diagram of CVS model.

The governing equation for pressure in the left and right ventricles is given by:

$$P(V,t) = e(t)E_{es}(V - V_d) + (1 - e(t))A(e^{\lambda(V - V_0)} - 1) \tag{1}$$

where E_{es}, V_d, A, λ and V_0 are constant parameters, and $e(t)$ is the driver function that varies in value between 1 and 0. Depending on whether inertia (inductors) is included in the flow between chambers, flow rate is governed by either of:

$$Q = \frac{P_{out} - P_{in}}{R} \qquad \frac{dQ}{dt} = \frac{P_{in} - P_{out} - QR}{L} \tag{2}$$

where P_{in} and P_{out} are the upstream and downstream pressures. The flow rates are then used to calculate the rate of change of volume in the chambers.

$$\frac{dV}{dt} = Q_{in} - Q_{out} \tag{3}$$

These primary governing equations are solved with flow rates and volumes as state variables. Initial conditions for the state variables are either obtained by distributing the volume so that the pressure in each chamber is the same, or by using the state vector from the most previously run model.

Unconstrained non-linear optimisation is used to identify model parameters for specified target performance. The variables include the resistances, elastances, and parameters used to calculate pressure (E_{es}, A, λ, V_d and V_0). At each optimisation step convergence is assumed when both the left and right ventricle stroke volumes are equal, and the

pulmonary and systemic circulations are equal. The target outputs are normal human CVS properties from Guyton (1991), as displayed in Table 1. CVS function is measured both from the average and the magnitude of the oscillation of pressure and volume waves. For example, a normal human aortic pressure is 120 over 80mmHg, which means the aortic pressure average is 100mmHg and the magnitude is 40mmHg.

The function to be minimised is set as the sum of the squares of the difference between all target and model outputs. Limits on the range of specific variables are included in exponential terms as penalty functions. Initial values for the parameters were approximated manually by trial and error.

		Target Output	Model Output	Error
Average Volume (ml)	lv	80	80.1	0.1
	rv	80	79.9	0.1
Average Pressure (mmHg)	ao	100	99.9	0.1
	pa	16.5	14.0	2.5
	pu	3	4.9	1.9
	vc	0	-1.3	1.3
Magnitude of Volume (ml)	lv	70	70.0	0.0
	rv	70	70.0	0.0
Magnitude of Pressure	ao	40	39.8	0.2
	pa	17	18.4	1.4

Table 1, Target outputs compared to model outputs from optimisation.

Results and Discussion

The optimisation results are shown in Table 1, where the volumes are within 0.1ml of target values, and the pressures are within 2.5mmHg. The resulting simulation using the optimised parameters is shown in Figure 2. These results show the potential of this technique for accurately modelling patient specific CVS function. Medical staff will be able to take measured patient data and obtain a patient specific model to assist in diagnosing irregularities. For example, if a particularly high ventricular elastance is required to model a patient, it would imply that the patient has stiff heart walls. Alternatively, a high aortic resistance in the model could alert medical staff to the possibility of occluded arteries. The known effects of certain drugs could also be tested to determine their impact on CVS function and assist in choosing suitable treatments.
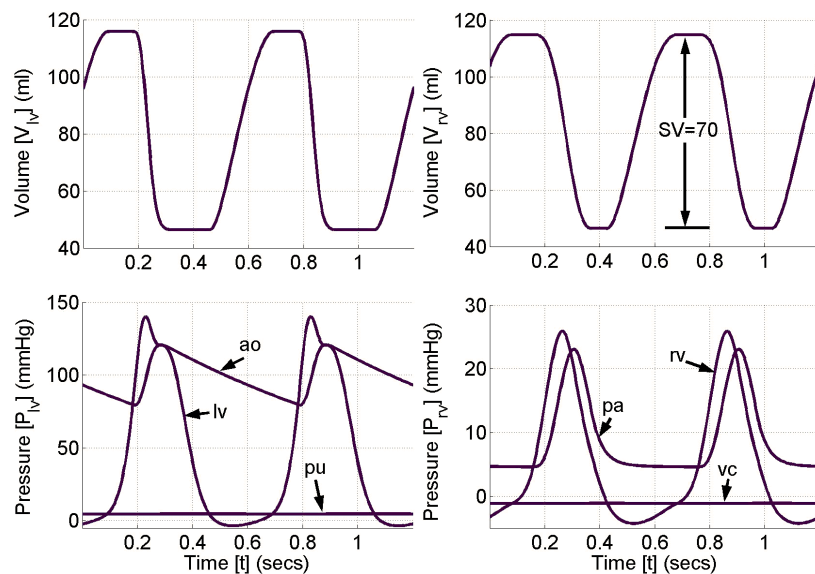
Figure 2, Plot of model outputs using optimised parameters.

Conclusion

While there are many examples in the literature of CVS models, there is little discussion on how to implement these models to assist medical staff in the diagnosis and treatment of patients with CVS dysfunction. Using optimisation techniques, it is shown that patient specific CVS models can be identified. Once identified, the patient specific model can assist in diagnosing cardiac dysfunction. This result offers the potential to bridge the gap between engineering models, and medical practice, combining the two to create software for practical patient diagnosis and treatment.

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