# Effect of Intra-aortic Balloon Pump on Coronary Blood Flow during different balloon cycles Support: A Computer Study

Thin Pa Pa Aye, Zwe Lin Htet, Thamvarit Singhavilai, and Phornphop Naiyanetr

*Abstract*—**Intra-aortic balloon pump (IABP) has been used in clinical treatment as a mechanical circulatory support device for patients with heart failure. A computer model is used to study the effect on coronary blood flow (CBF) with different balloon cycles under both normal and pathological conditions. The model of cardiovascular and IABP is developed by using MATLAB SIMULINK. The effect on coronary blood flow has been studied under both normal and pathological conditions using different balloon cycles (balloon off; 1:4; 1:2; 1:1). A pathological heart is implemented by reducing the left ventricular contractility. The result of this study shows that the rate of balloon cycles is related to the level of coronary blood flow.**

## I. INTRODUCTION

Nowadays, heart disease is the leading causes of death worldwide [1]. Therefore, with this general knowledge, prevention and treatment of the human cardiovascular system is critical. In recent years, mechanical circulatory supports as bridge to recovery, destination therapy, and bridge to transplant have been successfully used to manage patients with end-stage heart failure [2]. Among those patients who have utilized mechanical circulatory support devices, the intra-aortic balloon pump has been widely used to support heart failure patients because it is easy to apply, greater efficacy, safety, and cost effectiveness than other mechanical devices [3].

Understanding the interaction between cardiovascular system and intra-aortic balloon pump is an important challenge to know IABP effectiveness [4-6]. To demonstrate the relationship between the hemodynamic effect and phasing of the intra-aortic balloon pump, computer models have been created to validated such methods from data collected through animal trails and also clinical research done in the past [7-8].

One of the hemodynamic effects of intra-aortic balloon pump is that it increases coronary blood flow. To study the effect of coronary blood flow, the intra-aortic balloon pump counterpulsation was implemented in a swine model under normal and reperfused myocardium conditions [9]. Furthermore, the effect of oxygen consumption and coronary blood flow during intra-aortic balloon pump support was described in a canine model [10].

A model based on both the understanding of the hemodynamics, and blood flow dynamics is important for the explanation of cardiovascular diseases including heart muscle dysfunction, dilated cardiomyopathy etc. Computer modeling of the human circulatory system can help us to understand these conditions. This particular area of research will also be useful in physiology and medicine to develop in research and diagnosis. Several mathematical models of the human cardiovascular system have been implemented by many researchers [11-12]. The model of cardiovascular system for normal and pathological condition was established under the influence of baroreflex and pericardium [11]. Moreover, an each of left heart and right heart was improved as a time varying elastance during the valsava maneuver [12].

The objective of this study is to evaluate the effect on coronary blood flow during intra-aortic balloon pump support under normal and pathological condition with different balloon cycles.

## II. METHODS

### A. Mathematical Models

Cardiovascular (CVS) model: The complete schematic diagram of the cardiovascular model including intra-aortic balloon pump is shown in Fig.1. The model of intra-aortic balloon pump is added in the descending aorta model. The CVS model contains right heart, left heart, systemic and pulmonary parts. It is implemented by using lumped parameter model that is comprised of resistance, inductance, capacitance and elastance. The architecture of cardiovascular model is based on the study of Sun et al. [11].

For the myocardial perfusion model, the intramyocardial pump model is used to implement the left coronary circulation model reported by Schreiner W et al. [13]. To combine the coronary circulation model into the cardiovascular model, the inflow of coronary circulation (arterial blood flow) is connected with the aorta part. Similarly, the outflow of coronary circulation (venous blood flows) is connected to the central venous return part (right atrium). The CVS model descriptions have been reported in the literatures [11-12].

The mathematical representations of adding coronary circulation system into cardiovascular model are described in follows.

In the systemic arterial circulation element,

$$\frac{dV_{aa}}{dt} = Q_{av} - Q_{aa} - Q_{Lca} \quad (1)$$

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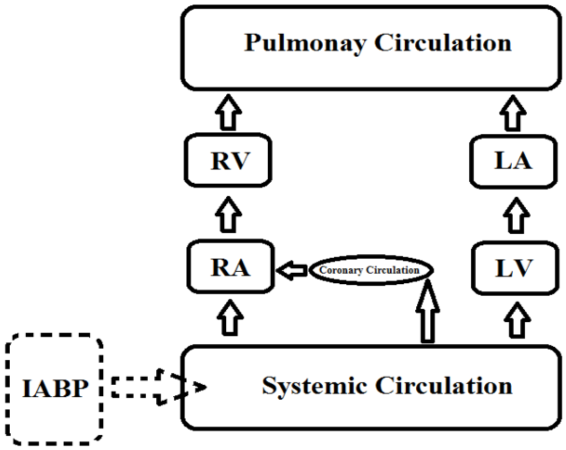


Figure 1. Schematic diagram of the cardiovascular system including coronary circulation assisting with intra-aortic balloon; RA: Right atrium, RV: Right ventricle, LA: Left atrium, LV: left ventricle, IABP: Intra-aortic balloon pump

In the systemic venous circulation element,

$$\frac{dV_{ra}}{dt} = Q_{vc} + Q_{Ven} + Q_{ex} - Q_{tv} \tag{2}$$

Where $V_{aa}$ is the ascending aortic volume (ml), $V_{ra}$ is the volume of right atrium (ml), $Q_{av}$ represents the flow through the aortic valve (ml/sec), $Q_{aa}$ represents the flow through the ascending aorta (ml/sec), $Q_{Lca}$ represents the left coronary arterial flow (ml/sec), $Q_{vc}$ means the flow through the vena cava (ml/sec), $Q_{tv}$ means the flow through the transcupid valve (ml/sec), $Q_{Ven}$ refers the flow through the coronary vein (ml/sec), $Q_{ex}$ refers the extraordinary venous outflow (ml/sec).

Intra-aortic balloon pump (IABP) model: The intra-aortic balloon pump is described by the balloon having a circular cross section with a vessel at any moment in time. The balloon is not distensible at the time when it is fully inflated and the volume of balloon is fixed when the balloon is deflated. A trigger of IABP inflation is detected from the systemic arterial pressure waveform. The starting time of balloon inflation is kept after the aortic valve closes (similarly with the dicrotic notch of the arterial pressure waveform) and the time of balloon deflation is set immediately before opening of aortic valve (similarly with the point just before the systolic arterial upstroke).

Physically, IABP is incorporated into the arterial section as shown in Fig.2. This intra-aortic balloon is placed into the descending thoracic aorta. Therefore, the volume of balloon is included into the volume of descending aorta. The mathematical equation is described by

$$\frac{dQ}{dt} = \left[ U_{in} \cdot \frac{dV_{in}}{dt} + P_{in} - Q.R - U_{out} \cdot \frac{dV_{out}}{dt} - P_{out} - P_B \right] / L \tag{3}$$

Where $P_{in}$ and $P_{out}$ are upstream and the downstream pressure of the compartment (mmHg), $U_{in}$ and $U_{out}$ represent viscoelastance of the vessel (mmHg s/ml), $V_{in}$ and $V_{out}$ denote volume in and volume out of the compartment (ml), $Q$ is the flow of the compartment (ml/s), R is the vessel

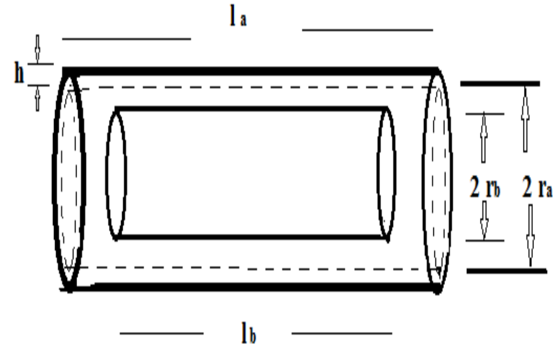


Figure 2. Balloon model in arterial section; h: thickness of arterial wall (cm), $l_a$: arterial length (cm), $l_b$: balloon length (cm), $r_a$: radius of aortic lumen (cm), $r_b$: radius of balloon (cm)

resistance (mmHg s/ml), L is the vessel inertance (mmHg s2/ml) and PB is the pressure of the balloon (mmHg).

The balloon compliance is the relation between the balloon pressure and volume of the balloon. So the balloon pressure is given as the following equation;

$$\Delta P_B = \frac{\Delta V_B}{C_B} \tag{4}$$

Where $\Delta V_B$ is the change in driving volume of the balloon (ml), $\Delta P_B$ is the change in pressure of the balloon (mmHg), $C_B$ means the capacitance of the balloon (ml/mmHg).

The resistance and capacitance of intra-aortic balloon are calculated according to the formula given as follows [5-7].

$$C_B = \frac{3 l_b \pi r_a^{\,3}}{2Yh} \tag{5}$$

And

$$R_B = \frac{81 \mu l_b}{8} \left[ \frac{4}{3} r_b (r_a - r_b)^3 + \frac{2}{3} (r_a - r_b)^4 \right]^{-1} \tag{6}$$

Where $l_b$ is the length of the balloon (cm), $r_a$ is the radius of arterial section (cm), Y means Young's modulus (N/m2), h is the wall thickness (cm), μ is the coefficient of viscosity (poise),$r_b$ means the radius of the balloon (cm) and $R_B$ is the balloon resistance (mmHg s/ml). The whole simulation model is validated with physiological books and published literature data [8,14 ].

*B. Simulation Study*

A mathematical model of the CVS including the coronary circulation assisting with the intra-aortic balloon pump is implemented by using MATLAB SIMULINK (MathWorks Inc, Natick, MA, USA. The complete descriptions of these models and theirs differential equations have been reported in literatures [11-13]. Parameters of the cardiovascular system including coronary circulation system are derived from literatures [11-13]. The CVS model is studied under both normal and heart-failure condition. The heart rate is

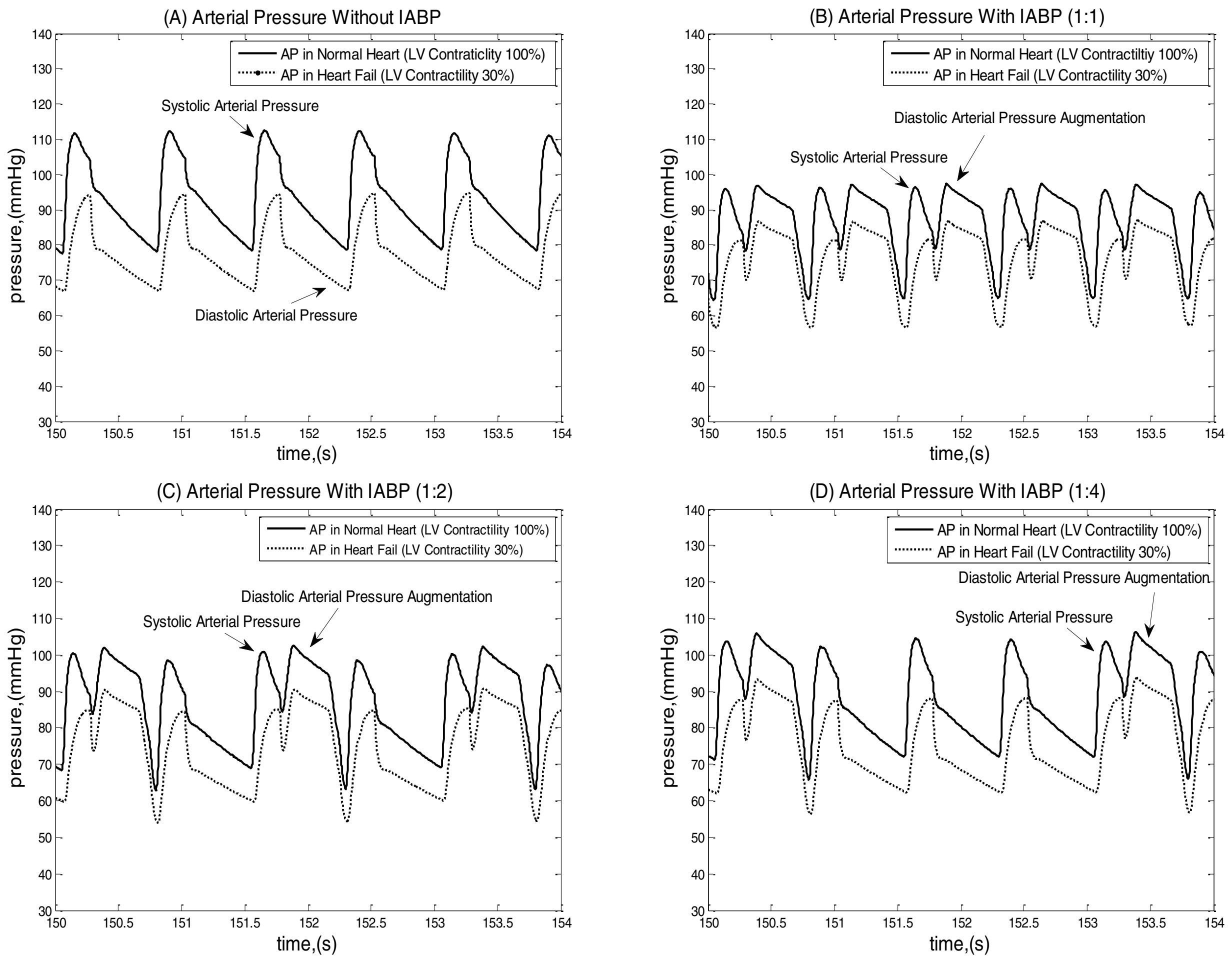


Figure 3. Arterial pressure waveform without IABP (A ) and with IABP, 1:1 (B) at normal heart and heart failure condition, AP : arterial pressure (mmHg) , Arterial pressure waveform with IABP, 1:2 (C) and with IABP, 1:4 (D) at normal heart and heart failure condition, AP : arterial pressure (mmHg) , LV contractility: left ventricular contractility.

maintained at (80) beats/min. The preload is set at control condition (EDV; end diastolic volume = 90 ml). The afterload is remained at constant. (SVR; systemic vascular resistance = 1.2 mmHg s/ml)

- For pathological conditions, the percentage of left ventricular contractility as represented Emax (maximum systolic elastance) is decreased. Emax is changed to 30% of normal conditions (Emax = 1.8 mmHg /ml).The volume of balloon is used as the driving volume for the pumping action of the balloon in this study. The balloon volume varies between 4 cc and 44 cc (drive 40 cc) at the deflation and inflation of the balloon, respectively. To set the transition period of the intra-aortic balloon pump, the balloon inflation and deflation is typically adjusted between 180ms [4].
- For different balloon conditions, the balloon cycles are studied when the balloon is in the following conditions: balloon off, 1:4, 1:2, 1:1 conditions. Balloon off means which it is not supported with IABP. 1:4 means one balloon cycle controlled in every four heartbeats. 1:2 represents one balloon cycle adjusted to every two heartbeats. 1:1 is each cycle set up per each heartbeat.

The hemodynamic effects of the intra-aortic balloon pump are studied for normal and heart-failure condition to compare with different balloon cycles.

## III. Results

### *A. Simulated Results*

The simulation results of normal heart and heart failure condition (reduced LV contractility 30% of normal condition) without supported IABP are shown in Fig. 3 (A). Arterial pressure in pathological condition is 15 mmHg lower than that of normal condition. The plot of hemodynamic simulated results in normal heart and heart failure condition (reduced left ventricular contractility 30% of normal condition) with IABP (1:1) are illustrated in Fig. 3 (B). Diastolic arterial pressure is augmented and systolic arterial pressure is decreased in both conditions. Similarly, the simulated results of arterial pressure with IABP on different balloon cycles (1:2,1:4) are shown in Fig.3 (C and D).

The simulated results of the left coronary arterial flow in normal heart and heart failure condition with and without supported IABP are shown in Fig.4. Left coronary arterial flow is increased during supported IABP in both conditions as shown in Fig.4. The cardiac output is increased during

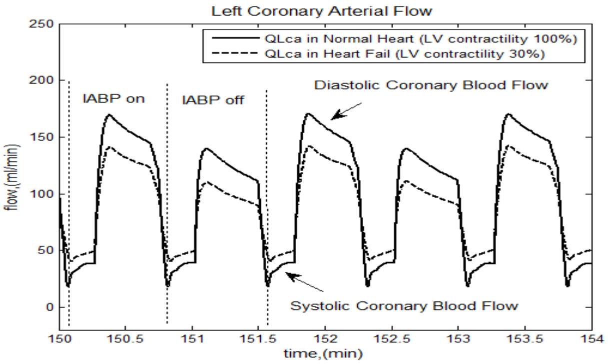


Figure 4. Arterial pressure waveform with IABP, 1:2 (upper panel) and with IABP, 1:4 (lower panel) at normal heart and heart failure condition, AP : arterial pressure (mmHg) , LV contractility: left ventricular contrictility

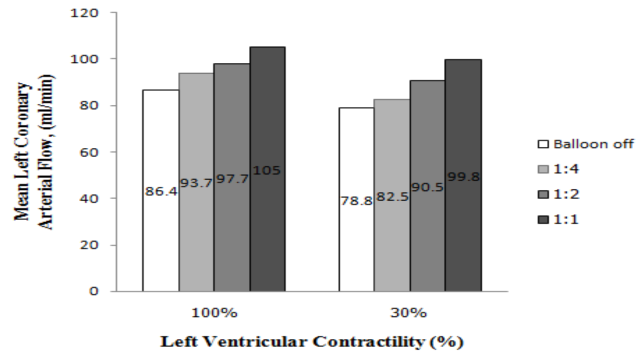


Figure 5. Mean left coronary arterial blood flow for normal heart and pathological heart conditions under different balloon cycles (balloon off, 1:4, 1:2, 1:1).

IABP support in both normal condition and pathological condition.

## IV. Discussion

The IABP mechanism used for cardiac support has been recognized to increase coronary blood flow by a diastolic arterial augmentation and reducing the left ventricle work load which is achieved by afterload reduction [3]. When assisting with IABP, the diastolic arterial pressure is agumented. This information is comparable to the patients data from previous literature [8].

During IABP support, systolic arterial pressure and left ventricular pressure are lower than those of non-IABP. Moreover, left coronary arterial blood flow with supported IABP is higher than that of left coronary arterial blood flow without supported IABP in both normal heart and pathological condition.

Under different balloon cycles, the mean left coronary arterial flow is progressively reduced from 1:1,1:2,1:4 and balloon off according to the assist ratio as shown in Fig.5. From this relationship, it is correctly confirmed with the patients data from literature [15].

The limition of this study is that the volume of IABP is fixed at 40cc and the balloon inflation and deflation is typically altered between 180 ms. Therefore, a study of adjustments of IABP timing and different balloon volumes would be further conducted in order to determine the IABP effectiveness.

## V. Conclusion

The computer study of the cardiovascular system and IABP under different balloon cycles has been conducted. The study provides the effects of hemodynamics during IABP support. This study should be useful in hypothesis development and serving a computer aided instructions concerning IABP support in clinical education.

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