

Chapter 7

Potential Cardiovascular Applications

7.1 Medical applications of the MIP

The valveless MIP is a simple design that offers a promising new technique for producing or amplifying a net flow for both macro- and micro scale devices without the requirements for valves or impeller. It may prove to be beneficial as an implanted device in the cardiovascular system to support or enhance the blood flow in specific vessels. Its simple design ensures a durable and reliable performance, as well as resistance to clogging due to the pulsatile nature of the flow and the lack of valves or blades. The pump performances may be controlled using modulations in frequency, pinching amplitude or pinching duty cycle.

We propose two direct applications. First, the MIP such as presented in chapter 2 could be directly used as an intra-aortic pump. It could serve as a bridge to heart transplant for patients suffering from congenital heart disease. Second, a MIP can be created by adding an actuator to a graft already coated with gelatin. This second application would have a great pediatric use for children suffering of an underdeveloped left ventricle.

7.1.1 Intra-aortic pump

Congestive Heart Failure (CHF) occurs when the heart is unable to pump blood effectively throughout the body. It accounts for 6% of the cardiovascular diseases, affects about 250,000 Americans each year, for who the mortality under medical therapy reaches 50%². In the most severe cases, when drug treatment is not sufficient, surgical therapy may include cardiac transplantation or implantation of artificial ventricular assist devices. Over the past 10 years, advances in drug treatment have helped CHF patients in all stages of systolic dysfunction.¹² Nevertheless, for a certain number, heart replacement is the only possible therapy.⁴⁵ This procedure is restricted by the limited number of donors, and registries show ever-longer wait lists. For these patients waiting up to 2 years for the surgery, there is a strong demand in a long-term minimally invasive circulatory support.

Mechanical assist devices can work by direct systolic augmentation of the heart, by mechanical pumping to divert blood from the left ventricle directly into the aorta with sufficient force to maintain normal arterial pressure, or by diastolic augmentation. In the later case, intra-aortic balloon pumps are the most commonly used but they require permanent hospitalization and can be left in place up to a few weeks only^{1,10,19,46,57}. In addition, complication rates range from 12% to 30 %¹⁶, and include limb ischemia, severe bleeding, balloon leak or failure. For long-term therapy, circulation need be assisted using mechanical assist devices such as left ventricular assist devices, bi-ventricular Pacers or total artificial heart. Left ventricular assist devices and total artificial hearts are electrically or pneumatically powered pumps that can produce continuous (Jarvik, DeBeckey) or pulsatile flow (Heart Mate, Novacor, Abiocr).^{12,33,39} They can be partially or totally implantable, but always require a power pack. Other major disadvantages

include heavy drug treatments to avoid thrombosis and to prevent infection.⁴⁸ In addition to a poor patient's quality of life, current long-term assist devices constitute an overall costly solution.¹⁸

An ideal cardiovascular pump³ should be reliable, durable, biocompatible, fully implanted, efficient, small in size, produce pulsatile flow, cause minimal thrombogenicity, and easy in placement and removal. The innovative MIP meets all these criteria, bringing intra-aortic pumps to a next generation. With the possibility of being inserted by a catheter through the femoral artery, the MIP will be positioned lying against the aortic walls. A gel-like liquid could be added *a posteriori* to recreate the thick gelatinous layer. The actuator can be made out of coils and magnets or a pneumatic chamber, and would be remotely controlled. The pump has many advantages as a cardiovascular assist device. It is fully implantable and controlled externally. It has barely any outward radial displacement when in use, and therefore would not strain the aorta. This low cost pump has no component that could obstruct the flow, damage the red blood cells and has minimal thrombogenicity. Finally due to its simple design, it is adaptable to each patient's morphology, and its use can be expanded beyond the case of heart failure.

7.1.2 *Gelatin-coated graft*

Hypoplastic Left Heart Syndrome (HLHS) is a fatal congenital cardiac malformation in newborns, in which the left side of the heart is underdeveloped resulting in a decrease in blood supplied to the body. Without treatment, 95% of these infants die during the first month of life, and none survive beyond four months.^{7,8} When possible, bypassing of the left ventricle following the Total Cavo-Pulmonary Connection (TCPC) Fontan surgical

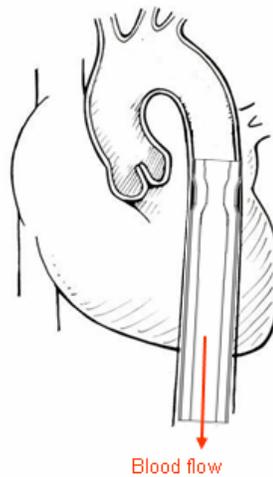


Figure 27. Illustration of the MIP fitting the inside of the aorta.

procedure is achieved by connecting the superior vena cava to the left pulmonary artery. As a result of this procedure, the single ventricle perfuses both the systemic and pulmonary circulations in series. Nevertheless, in many cases after the TCPC procedure the flow to the pulmonary system is not sufficient to ensure adequate pulmonary blood flow. In other cases severe conditions of systemic venous hypertension or pulmonary arterial hypotension are created. Other complications are attributed to disturbed flow dynamics in the TCPC. Typical reported failure rate and operative mortality of TCPC procedure is 9%¹⁵, and long term mortality after 15 years reaches 50%.³

Assisted cavopulmonary blood flow, by substituting a mechanical pump as a pulmonary ventricle to overcome this pressure gradient, would improve cardiac output by restoring the circulation to one more closely resembling two-ventricle physiology.⁴⁷ Mechanical circulatory support for adult patients who suffer from Congestive Heart Failure (CHF) is now state of the art clinically. Few studies tried to apply these clinically used cardiac assist devices for pediatric use as well in TCPC circulation.^{43,54,55} They used

animal models to prove the improved hemodynamics of pump-supported versus passive Fontan circulation. It was shown that pump support could conceivably reverse poor hemodynamics and secondary organ failure in patients with failing TCPC circulation, could make these patients more successful transplant candidates, or could be used as destination therapy. Even though improvement was clearly proven in all these cases, the pump chosen in all these studies was not suitable enough.

An ideal pump for cavopulmonary assist would require several features including pulsatile, net unidirectional flow, minimal thrombogenicity, and minimally invasive methods for placement and removal. The anatomy of a total cavopulmonary connection is favorable for circulatory assistance in that it provides a relatively straight longitudinal axis for instrumentation. The duration of assist needed for the univentricular support would presumably depend on patient-specific physiologic variables.

The MIP can serve as a noninvasive pump to enhance blood flow in the TCPC shunt to increase pulmonary flow and pressure independently of the systemic pressure (figure 28). The pump will actually be the original connecting graft with an added internal gelatin coating and an additional excitation device placed near its proximal edge. Its pumping mechanism would rely on the graft elasticity and the impedance mismatch at the graft connections. The MIP requires only small excitations to operate. In addition, the MIP has the great advantage of directly using the graft configuration. Impedance mismatch is achieved by the graft connections. One would only need to use a graft coated with a thicker gelatin layer than the already available Dacron gelatin-coated grafts. The pump is noninvasive and can be externally controlled. Because the pump performances are based on the frequency of excitation, it is patient specific and adaptable to patient

growth, by tuning the excitation frequency. The pump has, therefore, great potential for pediatric support, where demand of small, adaptable and noninvasive devices is especially important.

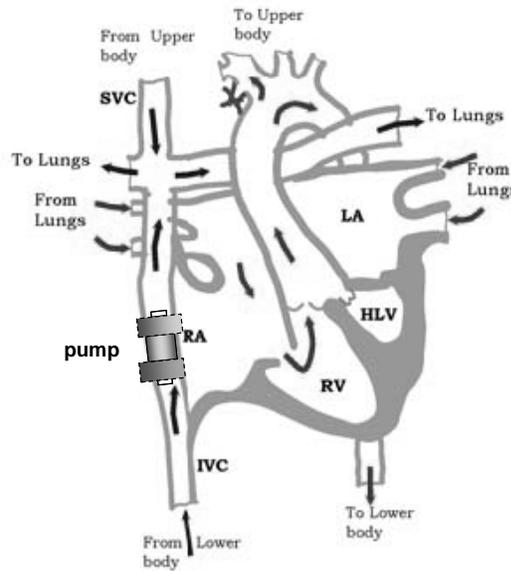


Figure 28. A schematic illustration of blood flow in the TCPC Fontan circulation.

(adapted from Migliavacca⁴⁴), with the pump.

7.2 Physiologically correct model of an aorta

To test the MIP as a cardiovascular device, the time-dependent boundary conditions at the entrance and exit of the device need to be carefully applied. To do so, we propose a simple boundary condition for modeling physiological flow and pressure in the descending aorta where the pump is to be implemented.

7.2.1 *Nature of the flow in the descending aorta*

The vertebrate circulatory system is a closed pressurized loop in which circulation is ensured by a muscular pump, the heart. During systole (40% of the cardiac cycle) the

heart contracts and ejects blood into the aorta. During diastole (remaining 60% of the cardiac cycle) the heart relaxes and fills with blood. During that phase, the aortic valve prevents blood flowing into the aorta. The arterial tree irrigated by the aorta provides resistance to the circulatory system, while the aorta serves as a compliance chamber. Over one cardiac cycle, blood flow exiting the heart is delivered in the aorta in an unsteady pulsatile manner, and so is pressure (figure 29).³⁶

The entire cardiovascular system may be simplified using a lumped parameter or one-dimensional (1D) model of flow. The earliest model characterizing the interaction between a pumping heart and systemic arterial tree was quantified by Otto Frank, and came to be known as the Windkessel model. It has been used to explain the rapid rise and gradual decrease of the flow and pressure waveforms.

The Windkessel model describes the cardiovascular system in terms of a compliant section in series with a resistive section. During systole, the compliant aorta acts like a capacitor to store blood. During diastole, the elastic aorta discharges the stored blood through the resistive branches of the smaller arteries to various organs. The pressure and flow waveforms given by this model are quite close to those measured in the body. Other investigators have produced more elaborate models with many elements in order to refine the waveform predictions.⁵⁰

The Windkessel model is here used as an outlet boundary condition to model pulsatile flow and pressure in the descending aorta. The proposed boundary condition model ensures the generation of a pressure waveform when pulsatile velocity inlet conditions are applied. The outlet boundary condition consists in an elastic vessel modeling the compliance chamber coupled with a porous media to ensure resistance (figure 30). The

aorta model with physiological boundary condition is to serve as a test bench for aortic devices.

7.2.2 Aorta model with pulsatile boundary conditions

The descending adult aorta is modeled as a straight tube of circular cross section ($R_{aorta}^f = 0.955$ cm, $L_{aorta} = 37.2$ cm).⁵⁰ The aortic tube is assumed to be consisted of a unique layer ($h_{aorta} = 0.075$ cm) of a purely elastic material within the range of biological soft tissue ($E_{aorta} = 65 \text{ e}+7$ dyn/cm², $\nu_{aorta} = 0.49$, $\rho_{aorta} = 1$ g/cm³). Blood is modeled as a Newtonian viscous fluid ($\mu_f = 0.035$ g/cm s, $\rho_f = 1$ g/cm³), and is governed by the Navier-Stokes equations (4) (5).

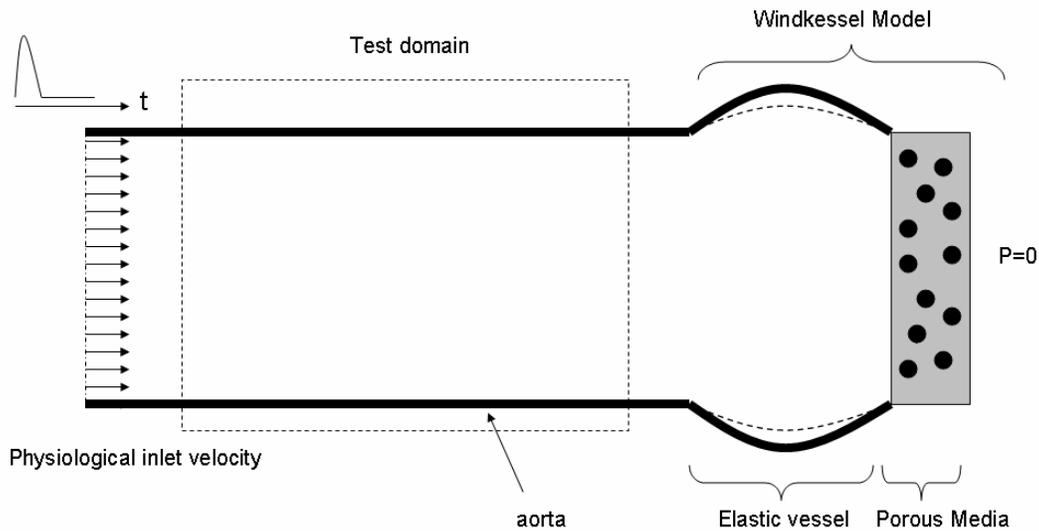


Figure 29. Schematic of the aorta model and the boundary conditions.

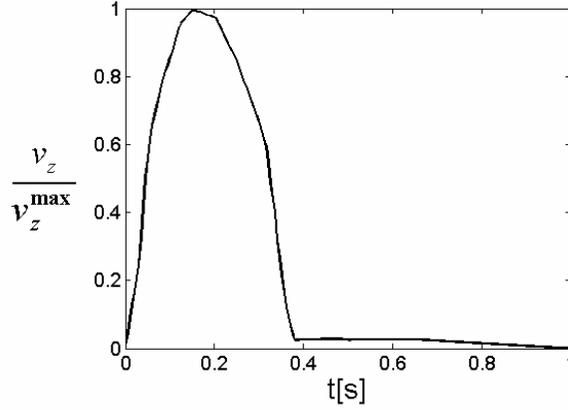


Figure 30. Imposed inlet velocity waveform. Plot of the axial velocity v_z is normalized by its peak value v_z^{\max} over one period ($f=1$ Hz).

The inlet boundary condition is an imposed flat velocity profile of pulsatile waveform (from Nichols&O'Rourke⁵⁰). The Windkessel model is situated at the outlet of the aortic tube. An elastic vessel ($E_{vessel}=1.5 \text{ e}+7 \text{ dyn/cm}^2$, $\nu_{vessel}=0.49$, $\rho_{vessel}=1 \text{ g/cm}^3$) is coupled with a porous media, which is a solid domain ($\rho_s=2 \text{ g/cm}^3$) saturated with the viscous fluid modeling blood (figure 31).

The flow in a porous media is governed by the Darcy's law:

$$\mu_f \boldsymbol{\kappa}^{-1} \cdot \mathbf{v} = -\nabla P + \mathbf{f}^b, \quad (34)$$

where μ_f is the dynamic viscosity, \mathbf{v} the velocity vector and $\boldsymbol{\kappa}$ the permeability tensor of the porous media, ∇P the pressure gradient along the porous media, and \mathbf{f}^b the external body force vector. The porosity ϕ , being the ratio of the volume occupied by the fluid to the volume of the mixed medium, was fixed to 0.8. The permeability tensor $\boldsymbol{\kappa}$ is assumed to have components in the direction of the flow only $\kappa_{zz}=2 \text{ e-} 4$.

The aortic walls and the elastic vessel walls are fully coupled to the fluid, and will be solved using the Fluid-Structure Interaction solver. The porous media walls are rigid.

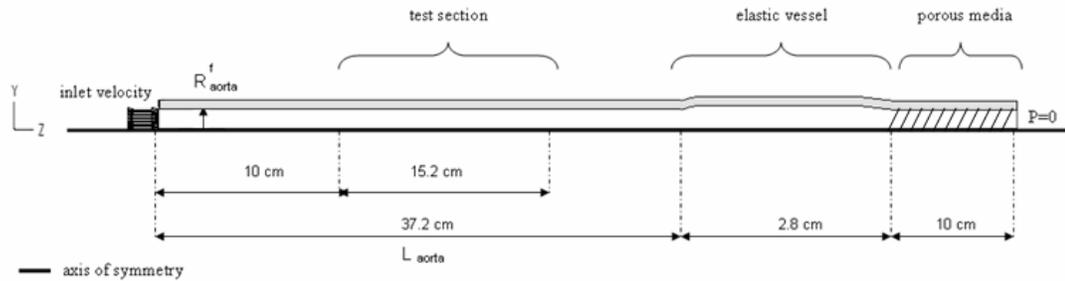


Figure 31. Physiologically correct model of aorta. 2D cross sectional view. Geometry and boundary conditions.

7.2.3 Results

Inlet pressure variation in time:

We fixed the inlet velocity magnitude (v_z^{\max}) to 50.5 cm/s so that to create a flow rate corresponding to the blood flow delivered by the heart of a sick patient (~ 2 L/min). The resulting pressure is found pulsatile. Only velocity was prescribed at the inlet boundary. The inlet pulsatile pressure is the result of the Windkessel model at the outlet of the tube. The pressure peak-to-peak value is consistent with the pressure peak-to-peak value found in a human descending aorta for elderly patient, slightly on the upper limit:⁴² $P_{\min} = 10 \text{ e}+4 \text{ dyn/cm}^2$, $P_{\max} = 21 \text{ e}+4 \text{ dyn/cm}^2$. Although the aortic walls have been modeled as fairly rigid, the pulsatile inlet flow generates waves in the aortic walls that are visible through the small fluctuations of the pressure waveform (figure 32).

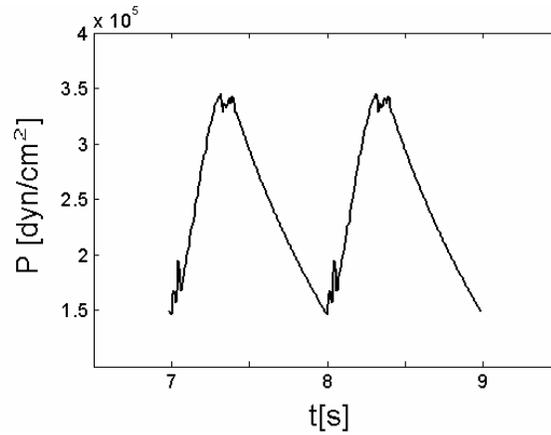


Figure 32. Resulting inlet axial pressure from the inlet imposed velocity profile.

Variation in time at the inlet for a node belonging to the axis of symmetry during 2 cycles.

Velocity and pressure variation at a cross section in the test domain:

In the test domain, the pressure and velocity waveform are conserved and of slightly higher values than to the incoming flow. They are still in physiological range and of waveform very similar to the one found in the descending aorta (The peak-to-peak pressure of a healthy middle-aged male patient is $125 / 78 \text{ mmHg} = 16.6 \text{ e}+4 / 10.4 \text{ e}+4 \text{ dyn/cm}^2$ and peak velocity is 50 cm/s).

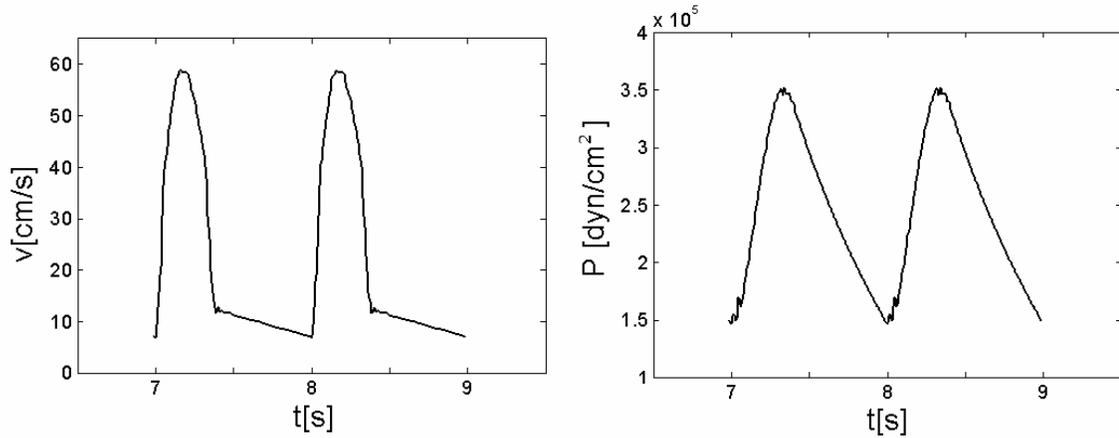


Figure 33. (Right) Axial velocity variation in time.(Left)axial pressure variation in time .

Plots are for the node located at mid length of the test domain on the axis of symmetry.

7.2.4 Summary

We propose a boundary condition that has the advantage of modeling the correct physiological pressure waveform when a pulsatile physiological velocity inlet condition is applied. The values of the flow rate and pressure are within the range of an elderly patient with a weak heart (aorta's wall modeled fairly rigid, weak mean flow rate, high blood pressure). This model can serve as a bench test for testing aortic devices. In appendix 3 a model of MIP inserted in the aorta with physiological boundary conditions is given.