Chapter 5

Impedance Pumping in the Embryonic Heart

The heart is the first functioning organ in the embryo. At very early stages of development the embryonic heat is a straight valveless tube for which pumping mechanism is still under debate. Fano and Badano[21] who, in 1890 were first to observe the embryonic heart contractile wave, characterized this primitive heart beating as peristaltic. Since their first observations, peristalsis has remained the accepted pumping mechanism. Recent investigations on impedance pumping[5,27] as well as *in vitro* visualizations of zebrafish embryonic beating heart[25] have invalidated this hypothesis and brought evidences that the embryonic heart may function as an impedance pump instead (section 1.1.2).

Using numerical simulations the two possible embryonic heart pumping mechanisms will be investigated. The exact same multilayered wall similar to the embryonic heart tube will be excited in two different manners to model peristalsis and impedance pumping in the embryonic heart. The multilayer tube is made out of an external stiffer layer similar to the myocardium layer and a soft internal layer similar to the cardiac jelly, and is the most physiological representative embryonic heart model to date. The multilayer tube is to be excited on its outer surface, the only layer in the embryonic heart containing contractile elements. The peristalsis model consists on the multilayer tube undergoing a sinusoidal wave of contraction along its length, whereas the impedance model uses a single excitation location such as introduced in section 2.2. Both models are excited at f=2Hz, the embryonic heart frequency, for an amplitude of 20% compatible with the myocardium shortening and both models pump a viscous fluid similar to blood.

For these conditions, results show that the peristaltic embryonic heart model produces a higher flow than the impedance one, but fails in building pressure, which is essential for the developing embryo. In addition, the peristaltic model requires more energy to actuate than impedance one. This expenditure is not compensated by the flow and pressure produced by the pump, resulting in an efficiency 5 times lower than the impedance one.

As a conclusion, compared to the embryonic heart peristaltic pump model, the impedance pump model seems to be best suited to provide in blood the developing embryo, while at the same time requiring minimum energy consumption.

5.1 The different embryonic heart pumping models

A peristaltic model for embryonic heart (PerisEHM) and an impedance embryonic heart (ImpEHM) were developed using the same multilayer tube introduced in section 2.2. They only differ by their mode of excitation (peristaltic wave vs. a unique excitation location). The same material properties for the stiff and the gelatin-like layer as well as the same boundary conditions (zero pressure at the exits and around the pump, no slip boundary condition at the fluid-structure interface, fixed ends at the tube's edges) to the one introduced in section 2.2 were applied. Both pumps featured the following specifics to mimic very closely the observed *in vitro* zebrafish heart dynamics[25]:

- the multilayer tube was filled by a viscous fluid similar to blood (μ_f =0.035 g/cm s, ρ_f =1 g/cm^3);
- the amplitude of excitation was 20% of the external radius, similar to the shortening resulting from the cardiac myocites contractions;
- the frequency of excitation was fixed to 2 Hz, similar to the one observed in the tubular heart.

5.1.1 The peristaltic excitation

The peristaltic excitation consisted in a wave of constant wavelength and constant amplitude traveling along the tube length (figure 19). The wave was imposed at the outer surface of the multilayer tube, on the layer that represents the myocites, the contractile elements of the heart tube. Because the extremities of the tube were fixed, the peristaltic wave was assumed to start and end at 1cm away from the edges of the pump. This way, unrealistic large stresses are avoided at the tube's extremities without impairing the physics of the peristaltic pumping. The peristaltic motion was practically modeled by prescribing sinusoidal radial displacements $\eta(t,z)$ on nodes belonging on the external surface of the external layer the tube (33). The amplitude of peristaltic excitation was set to 20% of the external radius (A=0.206 cm). The frequency of the peristaltic wave was f=2 Hz and the wavelength λ equaled the tube length L, so that the peristaltic wave was similar to the one observed *in vitro*[24] (figure 20).

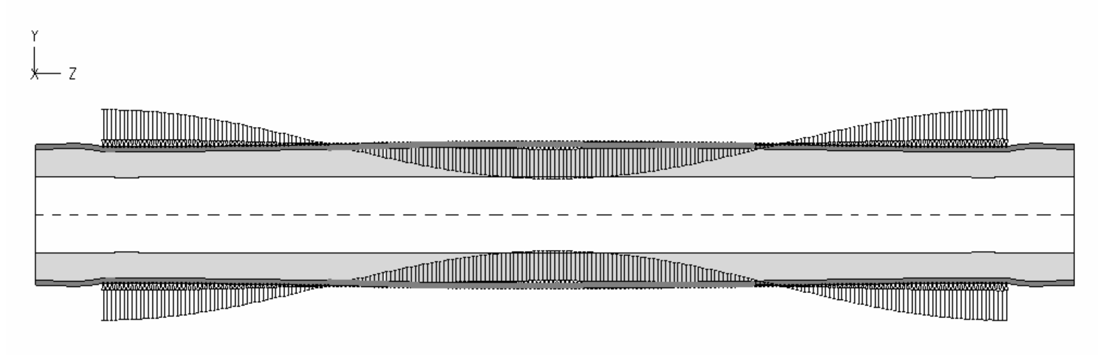

Figure 19. Model peristaltic embryonic heart pump. 2D longitudinal cross-sectional view of the multilayered tube with an imposed peristaltic displacement wave.

$$\eta(t,z) = A\mathbf{cos}\left[\frac{2\pi}{\lambda}(z-ct)\right], \qquad c = \lambda f \tag{33}$$

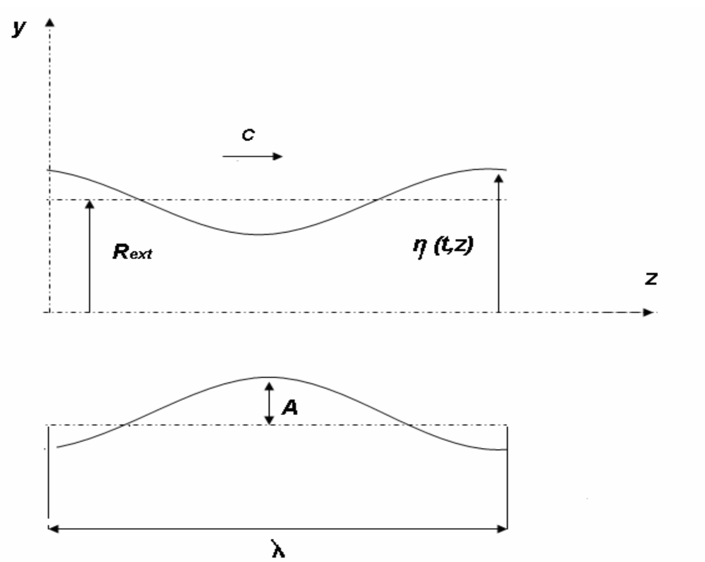

Figure 20. Schematic and conventions of the imposed peristaltic wave motion. *c* is the wave velocity.

5.1.2 The impedance excitation

The excitation of the ImpEHM was the exact same as the one described in introduced in section 2.1 (same pincher location, pincher width, pincher spatial distribution, and same duty cycle of 20%). Like in the peristaltic model, the frequency of excitation was fixed to f=2 Hz, and the amplitude of the compressive displacement was fixed to 20% of the tube's external radius.

5.2 Flow, pressure, and energy expenditure in the two models of embryonic heart pumping

5.2.1 Flow and pressure in the peristaltic pump model

Flow in the PerisEHM follows closely the imposed motion of the external walls. Although displacements are not directly imposed to the gelatinous layer, but to the external layer, the fluid-gelatin interface moves according to the imposed peristaltic wave, and no other wave motion is to be observed. The gelatin behaves as a passive layer transducing the peristaltic motion to the fluid domain. Its elasticity and soft constitution seems to not play a role in pumping. Exit flow is pulsatile and of the same frequency than the frequency of excitation, which is also the frequency of the traveling wave. Because the excitation amplitude is small and because the gelatin does not stretch significantly upon the passing of the contractile wave, the tube is never occluded. As expected from a partially occluded peristaltic pump[31] very little pressure is built up (figure 21). In addition, because the radius is never fully occluded, backflow is important. This type of peristaltic pumps which does not work by positive displacement; it relies on the viscosity of the fluid to entrain flow. The viscous forces alone can not ensure the pressure build up. The lack of pressure inside the pump combined with the large work expended to actuate the pump results in a poor efficiency of the peristaltic pump model (see table 3).

5.2.2 Flow and pressure in the impedance pump model

The ImpEHM behaves accordingly to the MIP flow and structure described in Chapter 3. After a phase of flow build up of about 10 cycles, the flow reaches a steady state of periodic oscillations. A large number of elastic waves are traveling and reflecting during the 80% of the period when the elastic tube is not compressed. These waves are responsible for the oscillatory exit flow and pressure waveforms (figure 22). Due to a constructive wave interaction, a suction-inertial expulsion mechanism characteristic of an impedance pump and such as described in section 3.9 drives the flow in a preferential direction. Because impedance pumps are kinematic pumps, they produce flow and pressure. A single excitation location is needed to actuate the impedance pump model, providing a small energy input for consequent output flow energy (see table 3).

5.2.3 Energy expenditure

The two pumping mechanisms, peristalsis and impedance, are actuated using prescribed displacement and the actuation work is calculated by integrating the nodal reaction forces over the nodal displacement and for a period of time (30). Because a zero pressure boundary condition is used at the pump extremities, the equivalent Poiseuille flow model introduced in section 3.13 is used to calculate the pumping work by using the equivalent Poiseuille pressure (32). The efficiency of each pump is defined as the ratio of pumping work over actuation work (29).

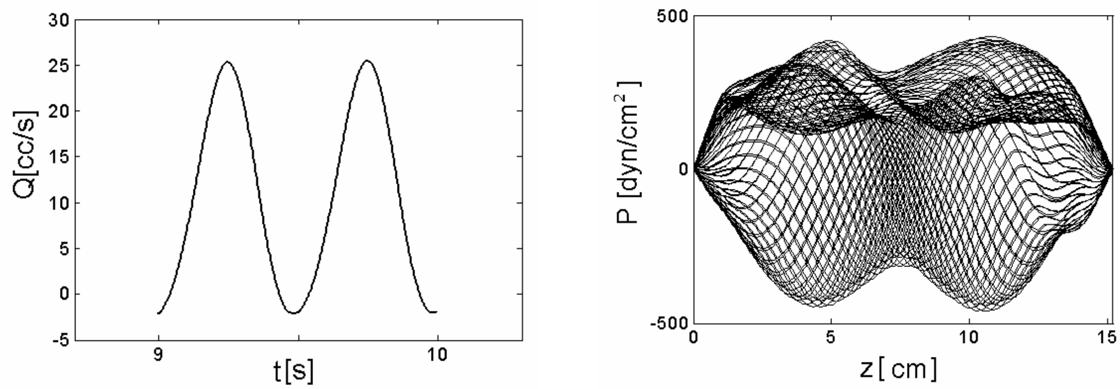

Figure 21. Flow and pressure in the PerisEHM. (Left) Exit flow rate and time. 2 cycles at periodic state are plotted. Mean exit flow is 11.87 cc/s. (Right) Axial pressure longitudinal distribution over 1 period. Each curve is the instantaneous axial pressure distribution in the tube.

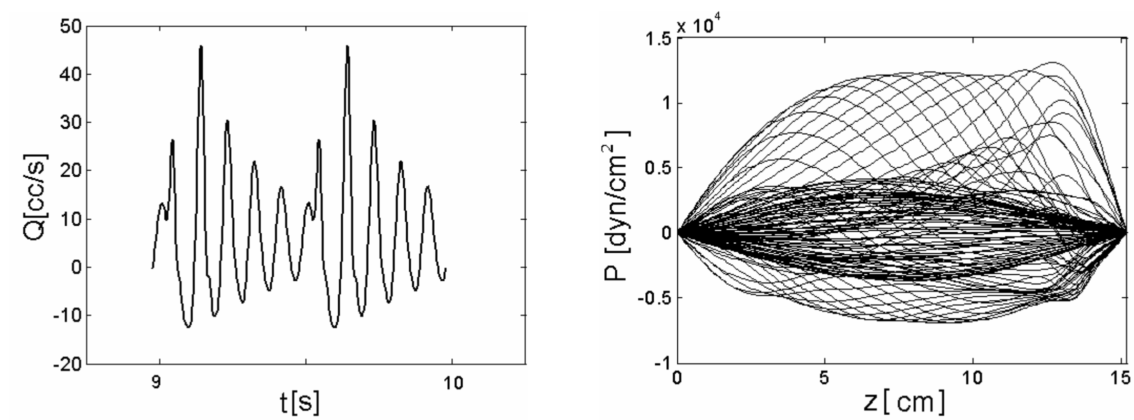

Figure 22. Flow and pressure in the ImpEHM. (Left) Exit flow rate and time. 2 cycles at periodic state are plotted. Mean exit flow is 6.20 cc/s. (Right) Axial pressure longitudinal distribution over 1 period. Each curve is the instantaneous axial pressure distribution in the tube.

Table 3 provides a list of the flow and energy characteristics of each pump model. Flow characteristics are the mean exit flow and the range of pressure observed inside each pump (figure 21 and 22). The peak to peak pressure is a quantitative tool for estimating the pressure that would be present at each pump's exit if the zero pressure boundary condition was not present.

Table 3. Comparison of the flow, pressure range, pumping work, actuation work, and efficiency between the multilayer impedance and the peristaltic heart pump models

	PerisEHM	ImpEHM
Exit flow	11.87 cc/s	6.20 cc/s
Axial Pressure Peaks	+/- 4 e+2 dyn/cm^2	-0.6 e+4 dyn/cm^2 +1.25 e+4 dyn/cm^2
Pumping Work	3792 erg	1650 erg
Actuation Work	7.09 e+4 erg	6.17 e+3 erg
Efficiency	0.0535	0.2673

5.3 Discussion

We compared two gelatin-coated elastic tubes based on the embryonic heart structure, and excited it in two different manners as models of embryonic heart pumping. The first model, PerisEHM, featured a peristaltic wave traveling along the tube, while the second, ImpEHM, used a single periodic excitation location and pumping relied on elasticity and wave interaction.

Flow and dynamic behavior are compared. The flow in the impedance model is about 1.5 times lower than in the peristaltic model but the impedance model produces a pressure that is of an order of magnitude 25 times higher. The fact that the flow is higher in the PerisEHM is a not a direct consequence of the mode of pumping -peristalsis versus impedance-. Indeed, the ImpEHM flow performances are nonlinearly dependant on the frequency of excitation (see section 3.6), and 2 Hz may not be a natural or harmonic of the natural frequency of the impedance pump. The spectral analysis of the impulse response exit flow history of the ImpEHM was found to be very similar to the one presented in section 3.4, where 33 Hz was a dominant frequency. Therefore for the considered ImpEHM, a slightly different choice of materials properties so that 2 Hz is a natural or an harmonic of the natural frequency of the system, would lead to higher exit flows.

Pressure however, is not the result of the choice of materials, but is intrinsic to the mode of pumping. Because the walls are only partially closed, the P_EH behaves as a viscous pump. It relies on the viscosity of the fluid to drive flow, and only small pressure is being produced. The lack of pressure produced by the peristaltic pump model has strong physiological consequences. As the embryo develops, the systemic resistance increases and the heart needs to produce enough pressure to overcome this resistance in order to irrigate all parts of the body. A pump that produces flow and only small pressure such as the peristaltic embryonic pump model becomes less efficient in delivering flow as resistance increases. The impedance pump however, whose mechanism is based on suction is capable of building pressure to accompany blood flow.

The comparison between the two pump models revealed the energetic advantage of pumping by impedance. The peristaltic pump model is 5 times less efficient than its impedant counterpart. In the PerisEHM every section along the length of the tube is undergoing at every instant an imposed displacement. In the ImpEHM however, a single location is used and active compression occurs for only 20% of the period time.

Finally, comparison of the two pump models is done for 2 periods when the ImpEHM has reached a steady state in flow. For this time frame, because of the long viscous diffusion time, the PerisEHM may not have reached its steady state. Using the impulsively actuated wavy wall model as an approximation of the wavy gelatin-fluid interface (2nd Stokes problem) one can estimate the time for the motion of the peristaltic walls to diffuse and influence the whole flow (see appendix 4). Based on the criterion that fully developed flow is reached when the axial velocity reaches 99% of the wall velocity, one finds a diffusion time of about 74 hours. This results implies that fully developed flow and peak performances in the embryonic heart excited by a peristaltic wave may not physically occur, due to the time scale difference between the viscous diffusion (74h) and morphogenesis (the heart tube shape changes every hour).

The high viscous diffusion time and the low pressure produced by a peristaltic pump are strong evolutionary incentives, and suggest that the embryonic heart may use impedance as an energetically optimized pumping model compatible with the physiological changes of the shaping heart. Through the comparison of peristaltic and impedance pumping in a multilayer tube similar to the tubular heart, we bring an additional piece of evidence that the embryonic heart may function as an impedance pump and rather than a peristaltic pump as previously accepted.