Chapter 4

Discussion

The multilayer impedance pump offers a new design for valveless pumping in confined environments where only small excitations are possible. The choice of materials for the pump model presented here is the optimized result of a previous series of combinations between elastic layers of different properties, with the scope to produce the desire wave amplification feature. The presented pump is an interesting resonant system, in which flow is not linear with frequency.

Several assumptions have been made. First, each layer constituting the elastic tube has been modeled as a linear elastic material. In the real case the stiffer layer may be rubber-like and the inner gelatin layer may be a viscoelastic material. This would introduce additional non linearities and material dissipation. In appendix 3 a model of MIP using a viscoelastic gelatin is proposed. Second, the dynamics of each layer have been solved using the small displacement, small strain hypothesis. Although strains in the gelatin can reach 0.3, they are of transient nature and do not affect the linear hypothesis. The error in linearization found by performing the same simulation with the large strain assumption is less than 0.05% (see appendix 2). Third, the model has been solved assuming a laminar flow. For the fluid-filled elastic tube problem, one may want to characterize the flow as unsteady laminar or unsteady *smooth* flow. The Reynolds number calculated using the mean flow can be as high as 9,959, which for a steady flow in a pipe is past transitional. In pulsatile flow, however, the onset of turbulence occurs only momentarily within the oscillatory cycle, when flow and velocity reach their peaks.[61] Therefore, the zones of turbulence may not influence the overall function of the pump. Modeling turbulence for the transient deceleration phases in the pulsatile cycles would not bring more to the understanding of the function of the MIP, which is used as a proof of concept on multilayer pumping where.

The MIP is a complex system and presents several natural frequencies. One could make an analogy between the present system (two solids coupled with a fluid), and a system of several coupled pendula, each of them having a different length. Upon excitation, each pendulum will oscillate in a manner that will exhibit a dominant frequency as well as frequencies induced by the interactions with the neighboring pendula. The system as a whole will not necessarily lock into a single resonant frequency, but will have several resonant or dominant frequencies. These dominant frequencies will be expressed at different strengths depending on the pendulum (or observable) under consideration. In a similar manner, the MIP spectrum reveals several natural frequencies. For the observable considered (flow rate at the exit) the dominant frequency if f=33 Hz. We studied the pump response around the first natural frequency of the system (f_n=11 Hz). We excited the pump behavior around this natural frequency, as opposed to the higher natural frequencies (f=41 Hz, f=49 Hz and f=60 Hz) because the PSD's dominant frequency f_d=33 Hz was its harmonic. The choice to study the pump at the natural frequency as opposed to the dominant frequency has been motivated by the fact that single layer impedance pumps were exhibiting the strongest response at the first harmonic of the dominant frequency of the PSD.[5,27,42] In addition, in the gelatin-coated pump, there is a trade-off between pinching amplitude and excitation frequency that limits the range of frequency a specific pump model can be excited at. Due to the gelatin softness, responses to higher harmonics (f=22 Hz, f=33 Hz, f=44 Hz, f=55 Hz) or to higher natural frequencies (f=41 Hz, f=49 Hz and f=60 Hz) are possible but would require smaller pinching amplitudes (see appendix 3). Finally, a frequency shift of about 8% is observed between the resonant frequency (f_{res}=10.1 Hz) and the natural frequency (f_n=11 Hz). When periodically excited, the successive pinches enhance wall motion and corrupt wave propagation. This may produce a shift between the resonant of the system deduced from the system response to periodic excitation and the natural frequency calculated from the impulse response spectral analysis.

We periodically excited the pump, and showed that the pump can produce bi-directional flow depending on the excitation frequency. The ability to reverse flow direction by adjusting the frequency of excitations has been reported by several open and closed loop experimental setups.[13,28,29,51,52] Positive flow, i.e. flow exiting the pump from the extremity the farthest to the compression zone, is achieved for frequencies close to the resonant frequency (f=[9 Hz, 12 Hz]) and reaches maximum at resonance (f_{res}=10.1 Hz). Negative flow is observed at frequencies below the resonant frequency (f=[8 Hz, 9 Hz]). We then focused on the pump response around the resonant frequency, when net mean flow is positive. For that range of frequencies, the pump exhibits the largest inner wall motion. The relatively large waves at the fluid interface never occlude the fluid domain. The minimum fluid radius observed throughout the computations is R_f =0.37 cm. The great gelatin stretch at resonance is to be correlated to the highest mean exit positive flow. The wave interaction mechanism leading to pumping is similar to the one of a classic IP.[5] At resonance their constructive interaction creates a suction zone toward the end of the pump that eventually expels the fluid in a jet like manner.

By considering the mechanical work done on the fluid by the long portion of the elastic tube past the pincher, we are able to show that the elastic tube itself acts as a pump and not as a passive resistor. For frequencies around the resonant frequency, the mechanical work is positive (tube does work on the fluid) although no active component such as a pincher is present in that portion of the tube. Upon actuation, the energy used to compress the fluid-filled elastic tube is transmitted into the elastic tube to deform it and into the fluid to move the fluid particles. The elastic tube and the fluid are exchanging energy along the tube, and at a given point along the tube (characterized by Avrahami and Gharib[5] as the velocity node), the elastic energy is given back to the fluid and contributes to the pumping. In addition, the mechanical work is clearly maximal at resonance, highlighting the concept of resonant pumping, where most efficient energy transmission between the passive elastic tube and the fluid is achieved. The mechanical work for frequencies 9.7 Hz, 9.8 Hz, 11.5 Hz is found close to zero although mean flow is non zero. This apparent contradiction is explained by the way the pumping work is defined. Because the pumping work is expressed as the pressure and kinetic energy difference between the input and output of the CV, a zero pumping work means that no energy *increase* is observed between the input and the output of the CV.

We evaluated the efficiency of the pump using a Poiseuille model in order to account for the pressure at the boundaries. This method in a comparative tool and is not intended to give the exact efficiency of the pump for each excitation frequencies. The efficiency is found to depend non linearly on the frequency of excitation, reaching maximum at resonance, highlighting the concept of resonant pumping.

The MIP can be used for different pumping applications. By tuning material properties of the elastic layers, one can pump fluids of different viscosities. The MIP has interesting features that are especially suitable for many biomedical applications. It has a simple and compact design, and has no component such as blades or valves that could obstruct the flow. The multilayer structure limits all large wave motion to the fluid-gelatin interface with almost no external wall motion. Significant pumping (up to 5.16 L/min) is achieved for small excitation (10% external radius) and the pump offers the possibility of bidirectional pumping, or switch, depending on the excitation frequency. Biomedical applications at the macroscale include circulatory assist devices, and include polymer pumping for drug delivery at the micro- and nanoscale.