Design and Simulation of MEMS Based Piezoelectric Insulin Micro-Pump

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Abstract: One of the most effective treatments for diabetes type 1 and 2 is the administering Insulin. Single needle mechanical insulin pumps are heavy and painful. A generation of micro-needle based MEMS based drug delivery devices can be an excellent solution for insulin dosing. The stackable structure provides minimum dimensions and makes the final product to be patchable in any flat area of human skin and in combination with Micro-Needle Array; it provides a safe, painless and robust injection application.

The design of positive volumetric insulin pump is significantly a multiphysics problem where the volumetric change of the main pump chamber and the pumped fluid are directly coupled. We used COMSOL Multiphysics to investigate the performance of a MEMS based Insulin Micro-Pump with a Piezoelectric actuator pumping a viscous Newtonian fluid. The model captures the accumulated out-flow, the netflow or flow fluctuations based on deflection of piezoelectric actuator which moves with the diaphragm disk in positive-negative directions, to induce discharge pressures at microneedle array based on different input voltages and different exciting frequencies.

Three COMSOL modules; Structural Mechanic, Piezoelectric Device and Fluid-Structure Interaction were used to study the 2-D/3-D models of this MEMS based concept. Fully Coupling Physics provide real time relationship between different parameters and pump's outputs. Post processing and ODE also are used to create different required outputs.

Keywords: MEMS, Piezoelectric, Insulin Micro-Pump.

1. Introduction

The MEMS based positive displacement insulin micro-pump [1] has a Piezoelectric Actuator on top of a diaphragm membrane made from Silicone glass. Induced vibrations from PZT actuator create positive/negative volume in the pump's main chamber, which pull fluid from Inlet gate and pushe it toward outlet gate. The gating process is governed by two PDMS Flapper check valves that control the fluid direction from inlet toward outlet leading to micro-needles. A distributer connects outlet gate to micro-needles substrate, and finally the established discharge pressure pushes the fluid out of Silicone micro-needle to skin epidermis, right above dermis layer [2].

2. General Dimensions and Materials

The micro-pump is designed based on the minimum dosage requirement for diabeties patients. In general, approximately one-half of the TDD (Total Daily Dose) is administered as basal rate. For most patients, basal rates are in the range of 0.01 to 0.015 units per kg per hour. For example a 60 kg woman approximately needs 0.6 to 0.9 units per hour (0.116 µl/min). The basal rates are adjusted empirically based on glucose monitoring results [3].

The general dimensions (Figure 1) of the pump with substrate structure is up to 15 mm in diameter and 2-3 mm in height (thickness). The designed hollow micro-needles are 200 µm in length and 30 µm in diameter at the pitch of 500 µm [2].

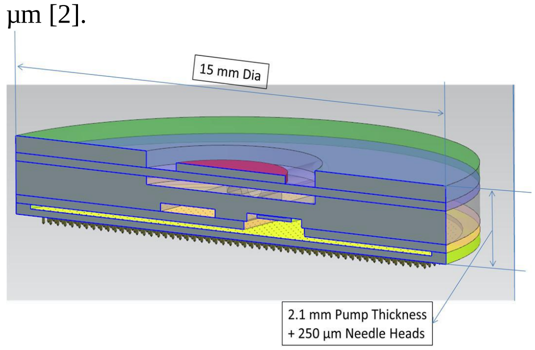

Figure 1. General dimensions and a cross section of Micro-Pump.

The main frame/structure material is a Silicone based solid. The Diaphragm disk is made of Silica glass. The Piezoelectric material is Lead Zirconate Titanate (PZT-5H). PDMS - Polydimethylsiloxane was used for check valve flappers. The Micro needle array substrate is made of solid Silicone through a deep-reactive ion etching (DRIE) manufacturing method [2]. Figure 2 shows the pump stack up layers. The main Si wafer as pump main body and micro-needle are fabricated based on MEMS technology and are epoxy glued to the rest of the layers [2].

3. Literature Review

There is almost no robust simulation analysis available for check valve based positive displacement micro-pumps with integrated micro-needles, other than using empirical analytical based on idealized flow and pump geometry. One of the biggest issue with flapper check valves is the simulation of moving mesh during solid contact in the gate closing process in FSI module. Although there is no mathematical solution available for this concept (due to topology change problem), however, we used a No-Contact but minimum gap to emulate the flapper function. It adds a very small leakage rate to the system. Because of the extreme non-linearity of the system, we used a fully coupled solver.

3. Governing Equations, Method, Boundary Conditions and Theory

The structural mechanics computations use the assumption that the material is linearly elastic, and only take geometric nonlinearities into account. The fluid flow is described by the single-phase, incompressible Navier-Stokes equations as this [4]:

$$\rho\frac{\partial \mathbf{u}}{\partial t} + \rho\mathbf{u}\cdot\nabla\mathbf{u} = -\nabla p + \nabla\cdot\mu(\nabla\mathbf{u} + (\nabla\mathbf{u})^T)$$

$$\nabla\cdot\mathbf{u} = 0$$

where **ρ** denotes the density (SI unit: kg/m^3), **u** the velocity (SI unit: m/s), **μ** the viscosity (SI unit: Pa·s), and **p** the pressure (SI unit: Pa). The equations are set up and solved inside the pump.

The Navier-Stokes equations are solved on a freely moving deformed mesh, which constitutes the fluid domain. The deformation of this mesh relative to the initial shape of the domain is computed using Hyperelastic smoothing. Inside the solid wall of the pump, the moving mesh follows the structural deformation.

Boundary Conditions at walls is no-slip:

$$\mathbf{U_{fluid} = 0}$$

For the fluid simulation, the boundary condition at the inlet and the outlet assumes that the total stress is zero, that is [4]:

$$\mathbf{n}\cdot[-pI + \mu(\nabla\mathbf{u} + (\nabla\mathbf{u})^T)] = \mathbf{0}$$

Boundry conditions at input:

$$\mathbf{n}^T\left[-p\mathbf{I} + \mu\left(\nabla\mathbf{u}_{fluid} + (\nabla\mathbf{u}_{fluid})^T\right)\right]\mathbf{n} = -\hat{p}_0$$

$$\hat{p}_0 \geq p_0, \quad \mathbf{u}_{fluid}\cdot\mathbf{t} = 0$$

And boundry conditions at output:

$$\mathbf{n}^T\left[-p\mathbf{I} + \mu\left(\nabla\mathbf{u}_{fluid} + (\nabla\mathbf{u}_{fluid})^T\right)\right]\mathbf{n} = -\hat{p}_0$$

$$\hat{p}_0 \leq p_0, \quad \mathbf{u}_{fluid}\cdot\mathbf{t} = 0$$

At the fluid-solid boundary, the structural velocity is transmitted to the fluid. As a feedback, the stresses in the fluid flow act as a loading on the inner boundary of the solid wall of the diaphragm.

Fluid-Solid interface boundry:

$$\mathbf{u}_{fluid} = \mathbf{u}_w$$

$$\mathbf{u}_w = \frac{\partial \mathbf{u}}{\partial t}, \quad \mathbf{u}_w = \mathrm{fsi.vWall}$$

$$\sigma\cdot\mathbf{n} = \Gamma\cdot\mathbf{n}, \quad \Gamma = \left[-p\mathbf{I} + \mu\left(\nabla\mathbf{u}_{fluid} + (\nabla\mathbf{u}_{fluid})^T\right)\right]$$

where $\mathbf{u}_{fluid}$ and $\mathbf{u}_w$ are the velocity vector of fluid and the diaphragm wall, **u** is the displacement vector, **σ (Γ)** is the stress tensor, and **n** is the normal vector to the FSI boundary.

The model's dependent variables are the displacements of the diaphragm wall together with the fluid velocity $\mathbf{u}_{\text{fluid}} = (\mathbf{u}_{\text{fluid}}, \mathbf{v}_{\text{fluid}})$ and pressure **p**.

To get the volumetric flow rate of the fluid $\dot{V}$ in m^3/s and the total volume of pumped fluid, we needed to perform some additional calculations. To obtain the volumetric flow rate at any instant t, we computed a boundary integral over the pump's inlet and outlet boundary. For round section:

$$\dot{V}_{\text{in}} = -\int_{s_{\text{in}}} 2\pi x(\mathbf{n}\cdot\mathbf{u})ds$$

$$\dot{V}_{\text{out}} = \int_{s_{\text{out}}} 2\pi x(\mathbf{n}\cdot\mathbf{u})ds$$

And for pump's square section inlet and outlet boundray:

$$\dot{V}_{\text{in}} = -\int_{s_{\text{in}}} 2x(\mathbf{n}\cdot\mathbf{u})ds$$

$$\dot{V}_{\text{out}} = \int_{s_{\text{out}}} 2x(\mathbf{n}\cdot\mathbf{u})ds$$

where n is the outward-pointing unit normal of the boundary, u is the velocity vector, and s is the boundary length parameter, along which we integrate. In this particular model, the inlet and outlet boundaries are horizontal so $n\cdot u = n_x u + n_y v$ simplifies to v or $-v$ depending on the direction of the flow [4].

It is of interest to track how much fluid is conveyed through the outlet during a pumping cycle, this can be calculated as the following time integral:

$$V_{\text{pump}}(t) = \int_0^t \dot{V}_{\text{out}}\, dt'$$

To compute this integral, we specified the corresponding ODE in COMSOL Multiphysics:

$$\frac{dV_{\text{pump}}}{dt} = \dot{V}_{\text{out}}$$

with proper initial conditions; the software then will integrate this equation [4].

3. COMSOL Multiphysics, Model Setup

The positive displacement Micro-Pump model is a combination of 3 physics; Solid Mechanics (solid), Piezoelectric Device (pze) (Electrostatic (es)), fully coupled with a Fluid-Structure Interaction (fsi) module. A Laminar Navier-Stokes equation used to simulate this problem. You can see the 2-D layout of the pump in Figure 2.

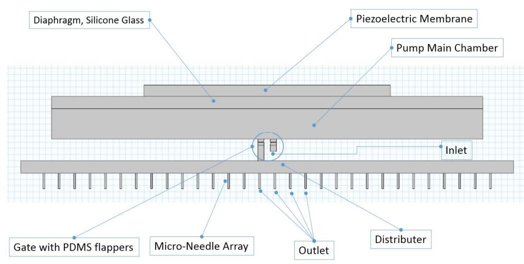

Figure 2. Micro-Pump model setup in COMSOL.

A wave signal at specific frequency excites the piezoelectric disk. Diaphragm disk and piezoelectric move together and FSI moving mesh transfers the stress/strain to fluid domain. Moving fluid-mesh follows solid deformation. FSI interface automatically handled by COMSOL Nonlinear models. With the action of flapper check valve, fluid flows from inlet to outlet. The fluid flow is described by the Navier-Stokes equations with laminar incompressible Newtonian flow and free boundaries at the inlet and outlet. The flapper material in check valves is PDMS which is suitable for a low stress/strain but high cycle applications. We used a standard Hyperelastic material model for this application. At the end, a Two-Way, fully coupled solver was used to calculate the results.

Integration coupling variables are used to track the discharged fluid and the volume of the fluid inside the pump.

A non-slip FSI boundary is automatically set up along the inner wall of the pump. COMSOL handles the fluid structure interaction using an Arbitrary Lagrangian-Eulerian (ALE) formulation. This involves a Lagrangian framework for the solid and an Eulerian framework for the fluid. A moving mesh model is used to track the deformation of the fluid mesh. The two-way coupling is captured along the FSI boundary by the fluid applying forces on The solid, and the solid displacement imposing a moving wall boundary condition on the fluid [5]. These conditions can be expressed as below:

$$\mathbf{u}_{\mathrm{fluid}} = \mathbf{u}_{\mathrm{w}}$$

$$\mathbf{u}_{\mathrm{w}} = \frac{\partial \mathbf{u}}{\partial t}, \quad \mathbf{u}_{\mathrm{w}} = \mathrm{fsi.vWall}$$

$$\sigma \cdot \mathbf{n} = \Gamma \cdot \mathbf{n}, \quad \Gamma = \left[-p\mathbf{I} + \mu\left(\nabla \mathbf{u}_{\mathrm{fluid}} + (\nabla \mathbf{u}_{\mathrm{fluid}})^{T}\right)\right]$$

where $\mathbf{u}_{\mathrm{fluid}}$ and $\mathbf{u}_{\mathrm{w}}$ are the velocity vector, $\mathbf{u}$ is the displacement vector, $\boldsymbol{\sigma}$ $(\boldsymbol{\Gamma})$ is the stress tensor, and $\mathbf{n}$ is the normal vector to the FSI boundary. COMSOL solves for the fluid velocity at the FSI boundary as an independent field.
A 3-D micro-pump COMSOL model was used to validate 2-D results.

3. Results

The micro-pump COMSOL model used to study the behavior of this pump for different input voltages (10 to 110 volt) with different input exciting frequencies (1 to 3 Hz).

The achievable discharging pressure for each case with zero back-pressure was investigated. In the next step, different back-pressures at outlet, were applied to the system and the system responses were captured accordingly. Each case outflow rates including accumulated flow (total discharge volume) and the netflow (fluctuation volume) were measured. Figure 3 shows the inflow/outflow rates at 110 Volt and 1 Hz. In Figure 3, any flow above zero shows leakage for outlet curve and in the same way, any flow below zero shows leakage for inlet curve.

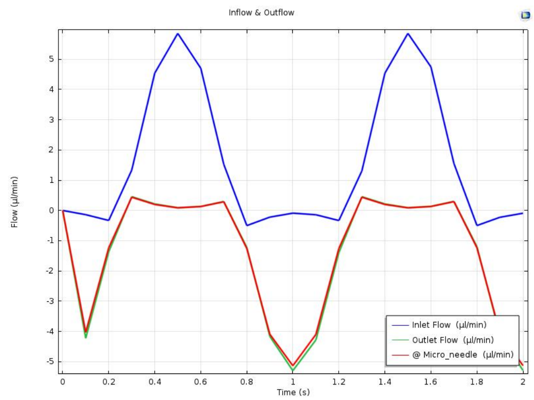

Figure 3. Inlet/Outlet Flows @ 110 Volt and 1 Hz.

Figure 4 shows netflow as accumulated flow or amount of flow fluctuation, and Vpump as volume conveyed or total discharge volume per time.

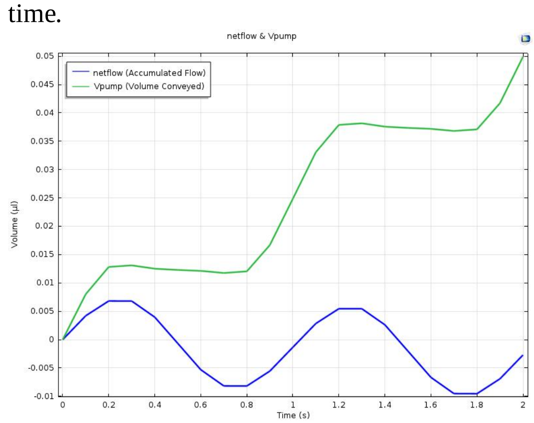

Figure 4. netflow and Vpump Flows @ 110 Volt and 1 Hz.

Figure 5 shows the stablished suction and discharge pressures at inlet and outlet. A negative pressure at inlet gate means suction pressure but the negative amount at outlet gate means discharge pressure.

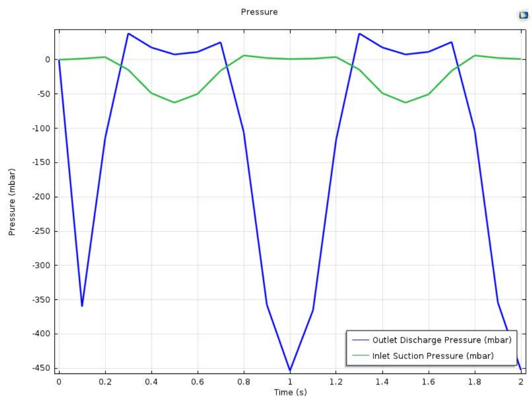

Figure 5. Discharge and Suction Pressures @ 110 Volt and 1 Hz.

Figure 6 shows Von Mises stress along with velocity magnitude in flapper check valves at 0.1 s.

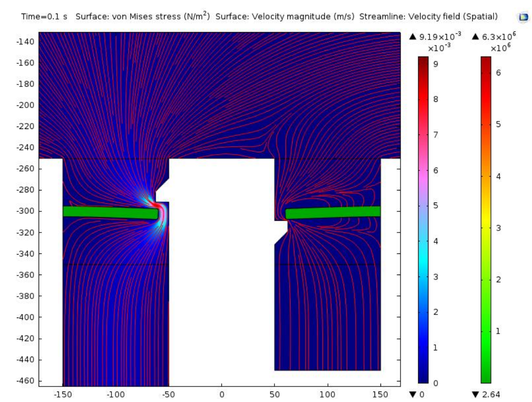

Figure 6. Von Mises Stress and Velocity Magnitude @ 110 Volt and 1 Hz in flapper check valves.

Figure 7 shows the diaphragm deflection at 110 Volt and 1 Hz. This deflection is a function of voltage and is independent of the exciting frequency.

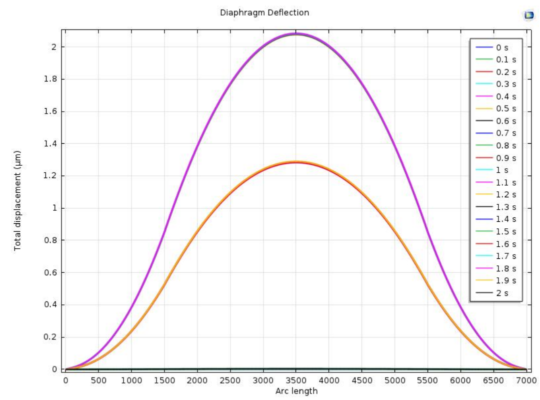

Figure 7. Diaphragm Deflection @ 110 Volt and 1 Hz.

Finally, Figure 8 shows amount of deflection of flappers in the check valves at 0.1 s.

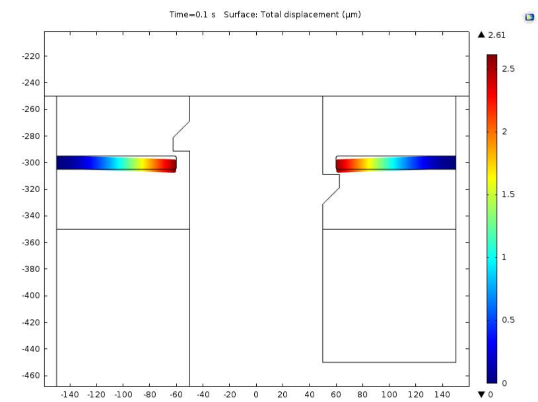

Figure 8. Check Valve Flappers deflection @ 110 Volt and 1 Hz.

All figures show the results for 110 Volt and 1 Hz input signal.

8. Conclusions

A parametric computational model of MEMS based Insulin Micro-Pumps was developed, using COMSOL Multiphysics. The model captures the pumping action based on the interaction between the piezoelectric, structure solid and fluid. It was used to study the pump performance under different inputs. The design seems acceptable for application of Insulin injection. The micro-needle device provides a safe and painless service to diabetes patients and performs well with this system. Mico-Pump performs correctly from the Min to Max spectrum of pressure and flow rates. The concept needs some support studies on durability and dynamic stability

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