Microchannel for Intraocular Pressure Relief and Glaucoma Treatment

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ABSTRACT

In this paper the deisgn, modeling, fabrication, and testing of an PDMS microchannel intended as an ocular implant for the maintanance of healthy intraocular pressure (IOP) and treatment of glaucoma are described. The winding serpentine microchannel provides a minimally invasive pathway for fluid to travel from the anterior chamber to the exterior of the eye. Analytical modeling using a friction factor correlated to channel cross section and FEM simulations predicted a pressure drop across the device around 2 kPa at the typical eye fluid production rate of 2 μL/min. A mm-scale and various μm-scale prototypes were fabricated by pouring and curing PDMS onto Su-8 molds patterned by photolithography. The pressure difference across the channel, which was designed to have a pressure drop of 2 kPa, was measured to be 1.47 kPa.

Keywords: glaucoma, PDMS, microchannel, MEMS

1 INTRODUCTION

1.1 Glaucoma

Glaucoma is an eye disease that leads to irreparable vision loss. Glaucoma is most often caused by high intraocular pressure (IOP), which in turn causes optic nerve damage. Normal IOP falls within the range of 1.33 – 2.8 kPa. Regardless of glaucoma type, its treatment (both pharmacological and surgical) is lowering IOP. IOP is maintained by a balance between the production and drainage of aqueous humor out of the anterior chamber of the eye, into the Schlemm's canal, and ultimately outside of the eye.

1.2 Glaucoma Drainage Devices

GDDs are implants that help to lower IOP by providing a pathway through which aqueous humor can effectively flow from the anterior chamber to outside of the eye. Many commercially available GDDs have a similar design, which has been modified very little in the last 50 years, in which aqueous humor is drained to a large plate sutured onto an outer layer of the eye. Newer designs have features that reduce complications of the plate designs. Unfortunately, neither design type properly addresses the flow regulation that must be sustained in order to effectively and consistently maintain a healthy IOP.

2 DESIGN

In order to treat glaucoma effectively, the implant must have the following features: an inlet ot outlet pressure differential of 2 kPa, a fluid flowrate equal to the eye fluid production rate of 2 μL/min, small size to minimize invasiveness, and biocompatibility.

During the design process, the ease of device fabrication was of great concern. Polymer MEMS and micromolding techniques are fast and inexpensive. This led to a simple, yet effective serpentine microchannel design shown in Figure 1 below.

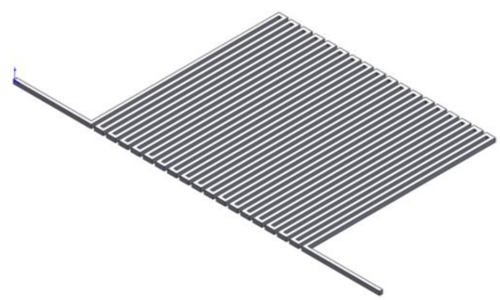

Figure 1: Isometric view of serpentine microchannel design.

Mask layouts for both mm-scale and μm-scale prototypes were made. The μm-scale prototype layout, shown in Figure 2, consists of channels with different widths, gap sizes, bend shapes, and total lengths.

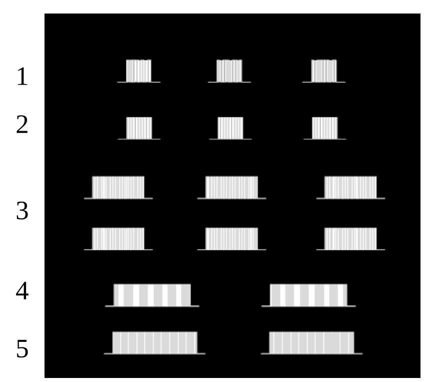

Figure 2: Mask layout of μm-scale prototypes. Rows: (1) 80 μm , 80 μm, and rounded turns, (2) rectangular turns, (3) 100 μm, 80 μm, and rounded turns, (4) 120 μm, 80 μm, and rectangular turns, (5) 120 μm, 100 μm.

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3 TECHNICAL ANALYSIS

Analytical and computer (FEM) models were developed for the microchannel design. Through these models the effect of various design parameters on pressure drop and flow rate were determined to estimate design performance.

3.1 Analytical Modeling

Due to the small fluid velocity and channel hydraulic diameter, the fluid flow in the channel is highly laminar, significantly reducing effects and losses assoiated with turbulent effects. The total pressure drop across the channel arises primarily from the friction with the walls.

In order to determine P_f, an empirical correlation between aspect ratio of the rectangular channel and the friction factor was used [1] [2]. The Darcy friction factor is given as follows: $f_D = C_1/Re$, where Re is the Reynolds number and C_1 is a constant, determined by experiment shown in the equations below.

$$P_f = C_1 \frac{QL\mu(1+a)^2}{8aA^2} \tag{1}$$

$$C_1 = 55.36 + 41.52e^{\frac{-3.4}{a}} \tag{2}$$

Where Q is the volumetric fluid flow rate, L is the total channel length, μ is fluid viscosity, A is the cross sectional area, and a is the cross section aspect ratio. Figure 3 is a plot of pressure drop as a function of flow rate at various cross sectional areas. A 40 cm long channel with an aspect ratio of 1 achieves a pressure drop of 2 kPa at 2 μL/min with a cross sectional area of 10^{-8} m^2.

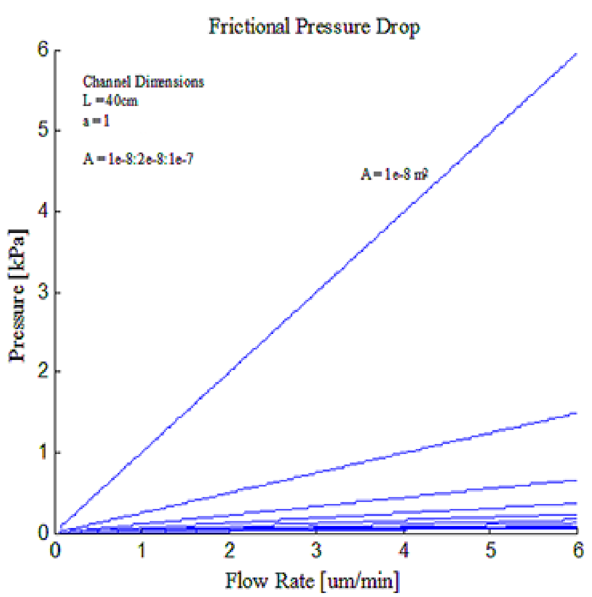

Figure 3: Frictional pressure drop across a microchannel at various flow rates and cross sectional areas.

3.2 FEM Simulation

Numerical models were run repeatedly for different dimensions and configurations of the microchannel. Results were compared both with the analytical models and the prototype testing performed later. The FEM modeling was especially useful for comparison with and troubleshooting of the analytical model. Figure 4 is of a channel with a width of 100 μm and a height of 170 μm. The overall dimensions are 5 mm by 10 mm. Results showed a pressure differential between the inlet and outlet of 2.7 kPa.

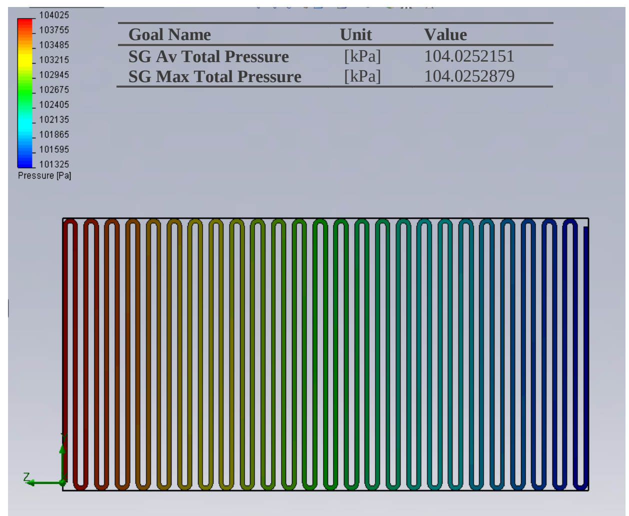

Figure 4: Solidworks Flowsimulation inlet pressure results for a 100 μm by170 μm microchannel.

4 FABRICATION

The device was fabricated by PDMS micromolding on an SU-8 mold. To make the mold, SU-8 was spin-coated onto a silicon wafer and patterned according to standard photolithography techniques to give a pattern with a thickness of 100 μm. PDMS was poured over the mold and cured. It was removed from the mold manually. Inlet and exit ports were punched in the channel, and the PDMS was bonded by oxygen plasma bonding either to another piece of PDMS or to glass.

5 CHARACTERIZATION

Two tests were performed to characterize the performance of the device. First, leak tests were performed to verify the success of the fabrication process. A syringe pump (KD Scientific) was used to flow water at a constant flow rate through the device. Any leaks could be observed. To determine the pressure drop across the microchannel, the syringe pump was connected to both the channel and a differential pressure sensor (Omega PX26). Priming the setup and connecting the tubing without large extraneous

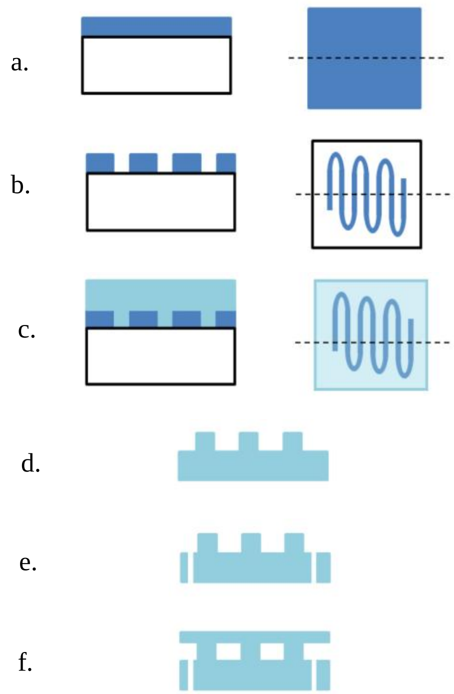

Figure 5: Fabrication Process: a. Spincoat SU-8. b. photolithography of SU-8. c. Pour PDMS. d. Demold PDMS. e. Punch Inlet and exit ports. f. Bond to another layer to close channel

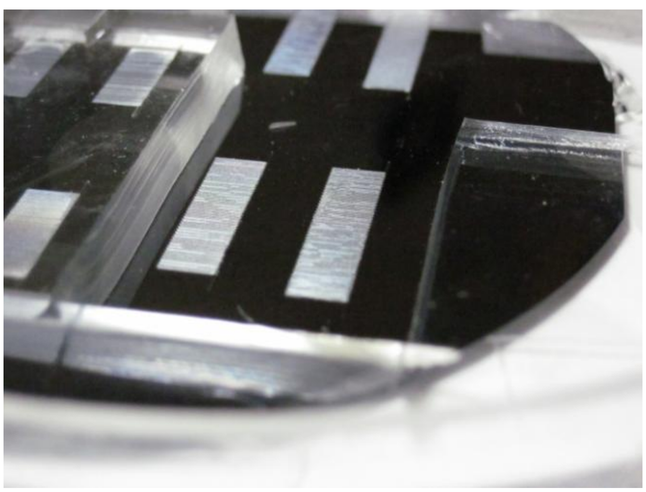

Figure 6: PDMS on SU-8 Mold

pressure readings proved problematic, so the sensor was tared after reaching steady state at a constant flow rate of 10 μL/min and the flow rate was increased in increments of 0.5 μL/min, with the steady state pressure recorded at each. Since pressure and flow rate are linearly related, this could be extrapolated to a flow rate of 2 μL/min. This method significantly reduced the noise and poor repeatability of the testing.

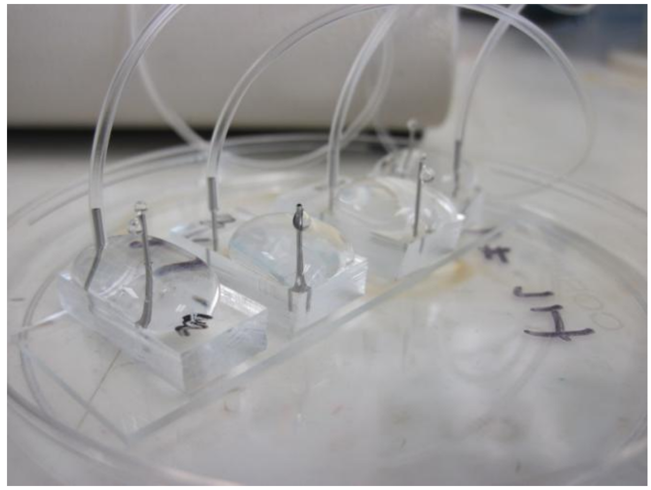

Figure 7: Leak Test

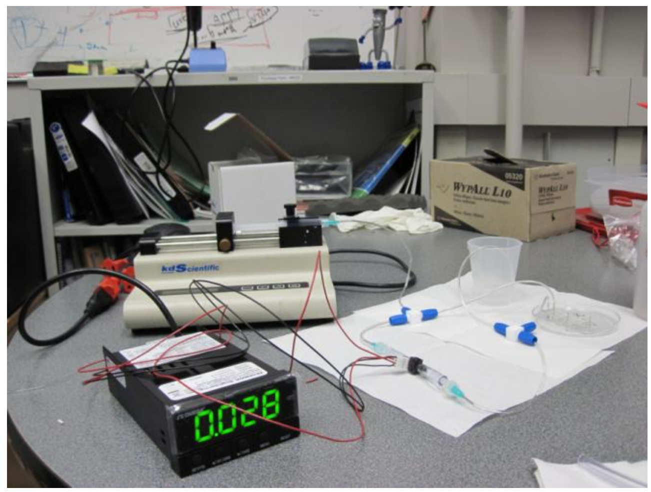

Figure 8: Testing Setup

6 RESULTS

In order to confirm the models developed and show proof-of-concept for the device, the pressure drop across the channel was measured and compared with these models. Initially, a device with milimeter-scale channel dimensions was fabricated for validation of the fabrication and testing methods. The pressure was measured at various flow rates and compared with both the analytical model and the finite element analysis. Results are shown in Table 1 below.

Flow Rate (mL/min)	Pressure (kPa)		
	SolidWorks Simulation	Analytical Model	Experiment
0.15	2.8	2.28	2.06
0.21	3.9	3.19	3.45

Table 1: Millimeter-scale prototype

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In testing of the microscale device, problems were encountered due to fluctuations of the pressure reading at zero flow rate. In order to eliminate these effects, the pressure was measured at incremental flow rates from 10 to 15 μL/min. As expected, the pressure change was linear with flow rate, so the results could be extrapolated to the desired flow rate as shown in Figure 9 below. The Pressure drop across the channel was determined to be 1.47 kPa, compared with 2 kPa expected from modeling. Much of this error can be explained by the difference in the actual size of the microchannel compared with its nominal size. The actual size has not been measured, but it is known to be larger than the nominal size.

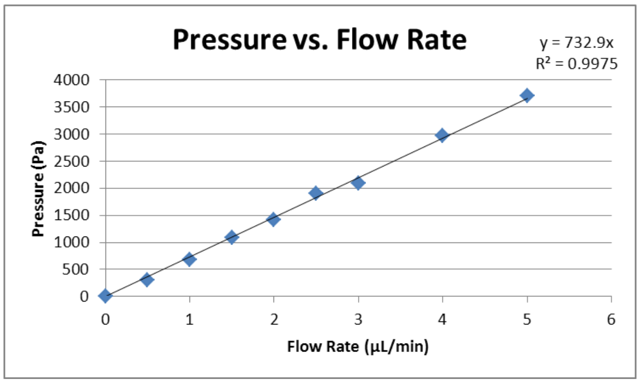

Figure 9: Determination of pressure drop

7 CONCLUSION

Modeling and testing confirmed the concept of using a MEMS fabricated serpentine microchannel made of PDMS as a glaucoma drainage device. The lower measured pressure compared to the models can be explained by the larger size of the channels than planned. This is most likely due to overexposure of the SU-8 during the mold fabrication. This could be fixed by more carefully controlling the photolithography parameters. Another posibility would be to use DRIE to make the mold, which may allow better control of the channel size. Better measurement techniques such as surface profilometry will be used in order to accurately measure the channels to make a better comparison between the model and the test results.

REFERENCES

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[2] W.M. Kays, M.E. Crawford, Convective Heat and Mass Transfer, McGraw-Hill, New York, 1980.