

A LOW POWER, MICROMACHINED, PROPORTIONAL VALVE FOR DRUG DELIVERY

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ABSTRACT

This paper describes a proportional valve that regulates viscous liquid media, intended for use with elastic reservoir/pumps in a portable drug delivery system. The valve is fabricated in a partially open state to permit flow even in the absence of applied power. At room temperature, piezoelectric actuation from -30 V to 60 V causes modulation of oil (viscosity 4.5 cP) flow rates in the range of 250-460 $\mu\text{l}/\text{min}$, while consuming as little as 0.136 μW of power. The valve is micromachined from bulk silicon and Pyrex glass. The entire packaged structure, including the actuator and housing, is 0.8 cm^3 .

Keyword: microvalve, piezoelectric actuation, drug delivery, flow modulation

1. INTRODUCTION

Elastic reservoir pumps are attractive for delivering liquids in portable applications since they do not consume any electrical power. However, the output varies with the fill level and delivery pressure [1]. These output variations can be corrected by using a proportional microvalve to regulate the pump. For applications that demand prolonged deployment and extended battery life, valves that consume little power are necessary. Using a normally open valve reduces energy consumption because it allows a nominal liquid flow rate without having to apply any power. This type of valve also provides greater control by allowing the flow rate to be either increased or decreased from its default liquid delivery rate. A number of micromachined valves have been reported in the past [5-14], but they have not addressed the needs of this particular application. Electrostatic valves typically require voltages >100 V, and are limited in operating pressure and range of flow modulation. Thermal and magnetic valves consume significant power. A piezoelectric valve used for this system greatly reduces energy consumption while overcoming the drawbacks of other actuation schemes. With proper control electronics and a coin cell battery, the entire portable system could interface with an individual through a catheter (Fig. 1).

Typical chemicals that might be delivered with a micropump include insulin, hormonal and pain management drugs, and any other liquid phase drugs that are preferably delivered in a gradual manner. (Table 1). These chemicals have a range of viscosities that must be taken into account in any portable delivery system design.

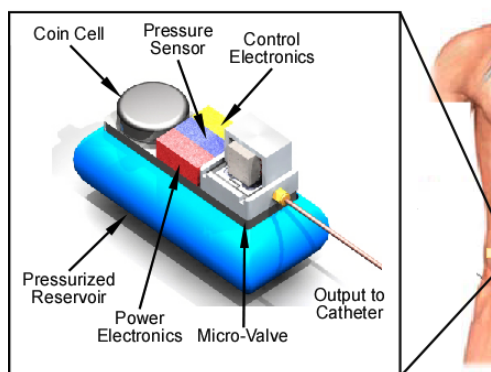


Fig. 1: Proposed system schematic. Expanded view of a side patch connected micro-valve controlled portable drug delivery system.

Table 1: Typical drug delivery flow rates.

Drug Name	Delivery Rate
Morphine	20-167 $\mu\text{l}/\text{min}$ [2]
Pethidine	10-250 $\mu\text{l}/\text{min}$ [2]
Epidural	0-200 $\mu\text{l}/\text{min}$ [3]
Insulin	2-350 $\mu\text{l}/\text{min}$ [4]

This paper presents preliminary investigations into using a normally open piezoelectric microvalve and an external passive reservoir [15] to demonstrate low power liquid flow modulation within the same ranges currently used to continuously deliver pain relief drugs such as morphine.

2. VALVE DESIGN AND FABRICATION

The valve is constructed from bulk silicon and glass, with a piezoelectric (PZT) stack actuator and a glass-mica (Macor™) ceramic cap. The PZT provides power-efficiency, high force, and proportional actuation. A buried oxide layer in an SOI wafer acts as an etch stop for deep reactive ion etching of silicon, so the valve flexures have uniform thickness. A recess is wet etched into a glass wafer to accommodate the PZT displacement. The inlet and outlet holes are formed using electrochemical discharge machining. The SOI and glass wafers are anodically bonded, diced, and prepared for assembly with the ceramic cap and PZT stack. To create a normally-open valve, the PZT stack is energized during the assembly process, so that it shortens after assembly. Finally, the finished valve is attached to a Macor header to interface with the passive reservoir/pump and standard tubing.

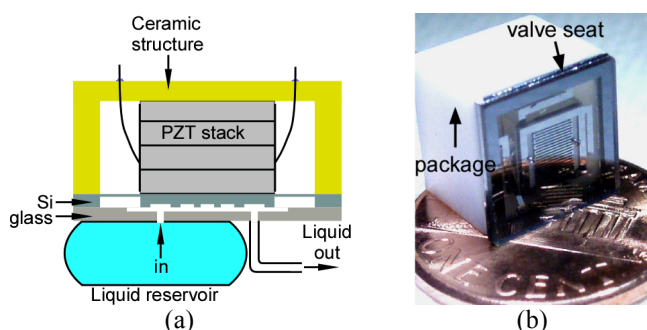


Fig. 2: (a) The valve cross-section showing the assembled ceramic-PZT-Si-glass structure and associated liquid reservoir. (b) A photograph of the completed assembled valve structure placed on a United States penny.

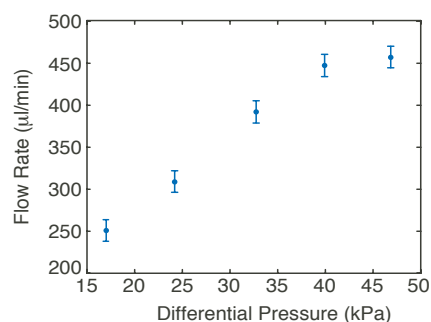


Fig. 3: Flow rate through un-actuated valve (0 V) across various pressures. At lower pressures the flow rate changes linearly with pressure.

3. VOLTAGE AND FREQUENCY FLOW RATE REGULATION

Preliminary tests were performed at room temperature using oil (viscosity 4.5 cP) to provide solution viscosity greater than saline water. For every test, a liquid reservoir was pressurized with regulated nitrogen gas to control and simulate varying pressures that might be generated from a passive source. The valve outlet was exposed to atmospheric pressure to create a pressure difference across the valve. Flow rates were determined by calculating the mass change in an outlet reservoir over a two minute time period.

In the first test (Fig. 3), the liquid flow rate was monitored as input pressure was varied. The flow rate varied linearly with the differential pressure over a range of 250-460 μl/min.; the corresponding differential pressure drop was 16-47 kPa. In the next experiment (Fig. 4), the valve was actuated by increasing and decreasing the voltage with different inlet pressures. Flow rates were tested over a range of input pressures (108-142 kPa), and could be altered from the un-powered, partially open flow rate by as much as 50 μl/min.

A frequency test was conducted in which the microvalve was actuated with a 50 Hz, 0-60 V square wave with varying duty cycle (Fig. 5). This provided more uniform linear control across inlet pressure than DC control. These results suggest that either voltage or frequency can be used to regulate the flow rate. DC operation consumed 0.136 μW and AC operation used 45 mW. As expected, DC regulation consumes less power. AC modulation is less efficient because of capacitive losses in the piezoelectric actuator.

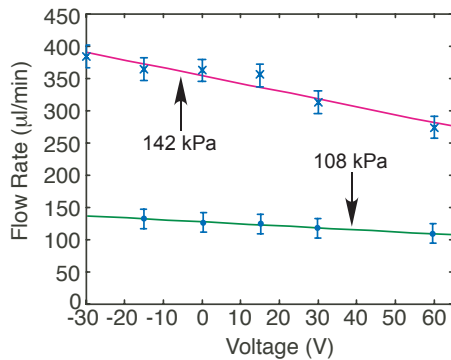


Fig. 4: Flow rate versus voltage actuation from -30 V to 60 V. Results are recorded for two inlet pressures, while the outlet is held constant at 101 kPa. Actuation is more pronounced with greater inlet pressure.

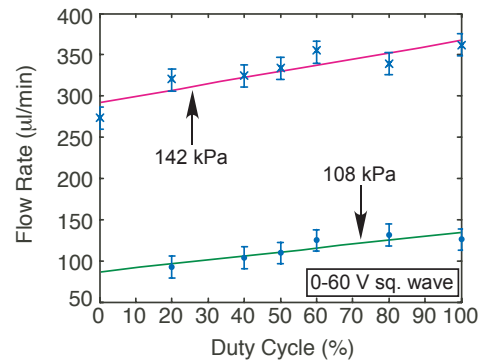


Fig. 5: Flow rate changes versus duty cycle for two inlet pressures. The actuation voltage was a 50 Hz square wave from 0-60 V. This method does not rely on the linearity of piezoelectric actuation.

4. CONCLUSIONS

The experiments described in this effort demonstrate that a piezoelectric microvalve actuated with varying DC voltages can adequately regulate liquid flow from a passive pump/reservoir. Furthermore, a power efficient, portable, high flow rate drug delivery system is realizable with these components.

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