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Normally-closed peristaltic micropump with re-usable actuator and disposable fluidic chip

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Abstract

This paper describes a piezostack actuated peristaltic micropump featuring a normally-closed function up to pressures of 100 kPa, when the electric power is off. The design is based on a modular setup with a re-usable actuator unit and a low-cost disposable fluidic chip that can easily be exchanged after contamination or use. Pump rates up to 40 µl/min at 28.6 Hz are demonstrated with water. The flow rates are backpressure independent up to 7 kPa, with a maximum backpressure of 45 kPa at 140 V. Also liquids with high viscosities up to 46 mPas are pumped successfully.

Keywords: Normally-closed, modular micropump, fluidic chip, piezostack micropump

1. Introduction

Increasing cost pressure in aging societies is one of the biggest challenges in the health care industry. One trend to cut down costs is to reduce hospitalization time per patient. Consequently, more and more ambulant (i.e. "out of hospital") types of therapy are required. For drug delivery for example, ambulatory infusion pumps are used to medicate people in their every day surrounding.

For a high level of security and flexibility, these pumps should offer the following features: i) *portability* (i.e small size and energy consumption), ii) *free flow prevention* ("normally-closed") when the electric power is off, iii) *modular setup* (i.e. re-usable pumping unit and disposable fluidic part), iv) *high precision*, v) *robust* liquid propulsion (e.g. backpressure independent) and vi) a *high flexibility* in terms of flow rates and flow rate profiles.

Although, many micropumps [1-3] and also some normally-closed micropumps [4-6] have been presented in the past, hardly any consistent portable pumping systems exists, that meets all the requirements listed above. The presented micropump is designed to meet these challenges and enable ambulant medication with a high level precision and security in the future.

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2. Design and Actuation

Fig. 1 shows the functional principle and a first prototype of the micropump, comprising a disposable fluidic chip and a re-usable actuator unit. The actuator unit itself can be sub-classified in an active part featuring the piezostack actuators and a passive part with spring elements. In the voltage free off-state ("normally-closed"), the fluidic chip is pushed against the plungers by the mechanical springs. Thus, the membrane is completely deflected and closes the in- and outflow opening at the bottom of the pump chamber.

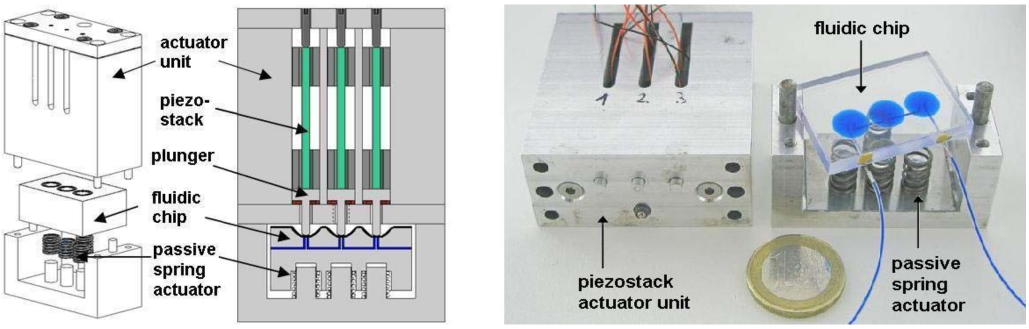


Fig. 1: Normally-closed peristaltic micropump. Left: Explosion and cross sectional view. Right: Prototype of the micropump with fluidic chip.

For pumping, all plungers are first moved downwards to their maximum deflection. This leads to a slight movement of the fluidic chip towards the passive spring actuators. Afterwards, the pumping sequence starts by moving the first plunger up and thus releasing the membrane (Fig. 2, left, 3). Consequently, liquid is drawn into the first chamber by the relaxing membrane. Subsequently, the fluid is transferred in the second and third pump chamber and finally out of the fluidic chip, resulting in an overall net flow (state 2-5 in fig. 2, left). Fig. 2 shows the electric drive signals (S1 to S3) of the three piezostack actuators to the corresponding fluidic situation within the chip.

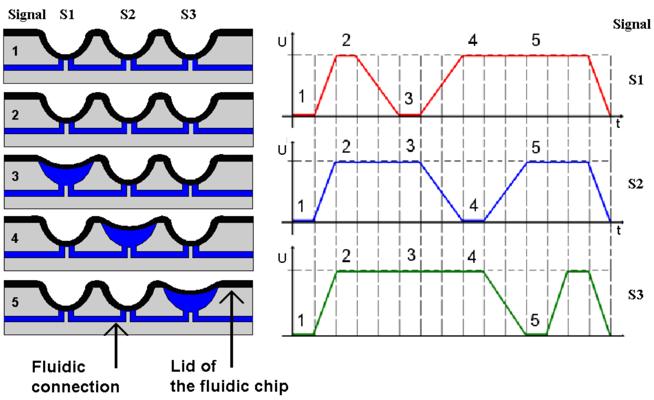
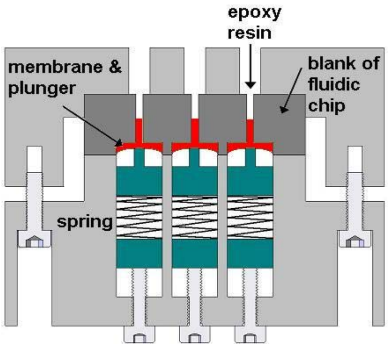


Fig. 2: Electric drive signals (right) and corresponding fluid motion in the fluidic chip (left).

3. Fabrication of the Fluidic Chip

The pump chamber bottom of the fluidic chip should adapt the shape of the maximum deflected membrane to enable a sufficient sealing at the in- and outlet and thus the desired normally-closed state. For this, the pump chambers are casted as a negative of the deflected membrane using an epoxy resin (Struers / SpeciFix-20). Therefore, the membrane is deflected in an assembly tool while the cavity underneath the membrane is filled with epoxy resin (fig. 3, left). For a proper release after the curing process, the surface of the membrane was pretreated with car polish. In the first prototype version of the fluidic chip, this process was done for every individual fluidic chip (fig. 3, right picture, left side). In the final version (fig. 3, right picture, right side) a master chip was casted in a mold and the data of a profilometer scan were used to micro-milling the pump chambers directly into a polycarbonate block. Additionally, the fluidic connections between the pump chambers are milled and holes for the in- and outlet connections are drilled into the chip. After gluing stainless steel tubes into the drilled in- and outlet holes as connectors for a flexible tube, the chip body is ready for the capping process of the lid. This is done in a heat bonding step with a laminator (Weiss & Soehne, Pilot Coater PCS-30) and a hot melt adhesive (Buehnen D 1544).



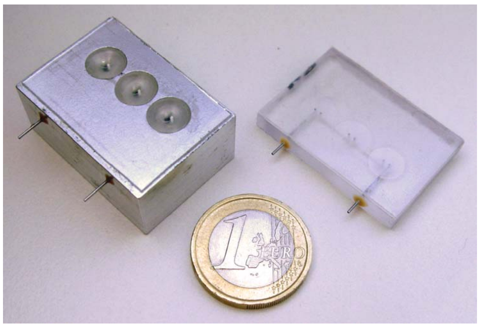
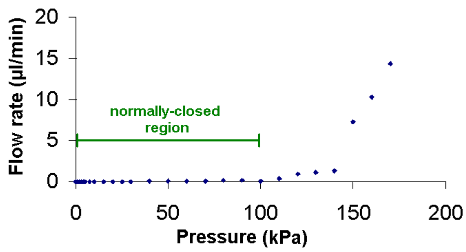


Fig. 3: Assembly tool (left) for first prototype (right picture, left side) and the succeeding version (right picture, right side) of the fluidic chip.

4. Experimental Results

The micropump is constructed for a precise dispensing of small volumes with flow rates between 0 – 100 µl/min. The characterization of the normally-closed state, using the first prototype of the fluidic chip (fig. 3, right picture, left side), is depicted in fig. 4 (left). Up to a pressure of 100 kPa, no fluid flow was measured. At higher inlet pressures, free-flow (i.e. a measurable flow rate) is observed. This proves a good sealing between the membrane and the in- and outlet in the pump chamber bottom by the force of the passive actuators, up to a pressure of 100 kPa. The backpressure behavior of the micropump can be seen in fig. 4 (right) for two different frequencies. A backpressure independent flow rate (flow rate above 90% of the maximum) was obtained up to critical pressure p_c of 7 kPa. The maximum pressure p_{max}, where the pump rate reaches zero because of the backpressure, is 45 kPa.



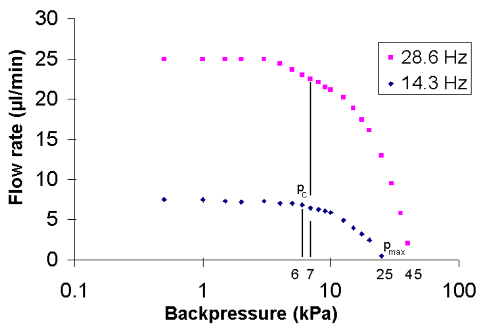


Fig. 4: Left: Test of the free-flow behaviour. No fluid flow was observed, up to a pressure of 100 kPa (normally-closed region). Right: Backpressure behaviour, the flow rate is independent of the backpressure up to a critical pressure p_c and afterwards linear dependent up to p_{max}.

The frequency behavior for different amplitudes is shown in fig. 5. The graph shows a typical behavior of a micropump. Up to a certain frequency (15 Hz) a linear correlation between flow rate and frequency exists. For higher frequencies, the flow rate reaches a maximum value of 40 µl/min (at about 30 Hz), before slowly declining again (not shown in fig. 5).

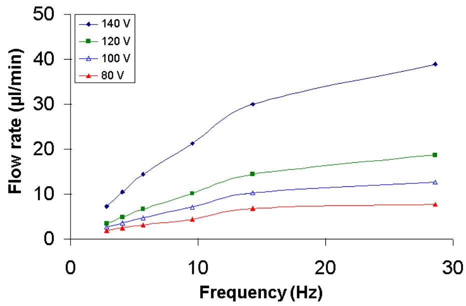


Fig. 5: Flow rate against pumping frequency for different voltage amplitudes of the piezo signal.

5. Conclusion

The presented piezostack actuated micropump features several relevant characteristics for ambulatory infusion therapy. The normally-closed mechanism in the off-state and the backpressure independent and stable flow rate are the most important ones. Additionally, the modular setup with a low-cost fluidic chip and a re-usable actuator unit perfectly fits the needs in the medical field. The next step towards a prototype for field use is the implementation and integration of the required electronics and energy supply to an integrated, portable system.

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