PipeJet™ - A SIMPLE DISPOSABLE DISPENSER FOR THE NANOLITER RANGE

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Abstract:
This paper reports for the first time on an unrivalled simple, disposable non-contact dispenser for the nanoliter range. In contrast to other known dispensers manufactured by silicon micromachining [1-4] the new device simply consists of an elastic polymer tube with circular cross section. Actuation is done by a piezostack driven piston, squeezing the tube at a defined position nearby the open end by a significant fraction of the cross section. In contrast to drop-on-demand devices based on an acoustic actuation principle [5] this leads to a significant mechanical displacement of the liquid. In our experiments we successfully tested ten media in the viscosity range from 1 mPas to 27 mPas and even up to approximately 200 mPas with a new actuation setup. Frequency characteristics showed an independent dosage volume for water up to a frequency of 15 Hz. Standard deviation within 1,000 shots resulted in an excellent CV of less than 2% of the dosage volume. With a print frequency of 340 Hz a flow rate up to 143 µl/s has been reached. Beyond the possibility to dispense highly viscous fluids also emulsion paints with particles of approximately 40 µm diameter have been printed successfully.

Keywords: non-contact dispenser, nanoliter range, low-cost, drop-on-demand, high-viscous fluids

Introduction

The control of smallest amounts of fluid is of increasing interest in drug development, modern diagnostics and chemistry. Handling liquid volumes down to a few nanoliters with high reliability and high accuracy is very important for the reduction of the assay volumes especially in High Throughput Screening applications. Compared to the conventional pipetting systems, where the precision of the dosage volumes below 1 µl is limited by capillary and adhesive forces, dosage systems with free flying liquid jets or drops have the arbitrative advantage to avoid these surface interactions. Because of the variety of different fluids used in chemical and biochemical laboratories with very different properties it is essential that the dispenser provides a dosage volume nearly independent of fluid properties like viscosity, surface tension or density. So that it is not necessary to calibrate the system after every used liquid. For medical devices it is also recommended that all contaminated parts are disposable and therefore low-cost and easy to handle. Furthermore there is an increasing demand for dispensers which can dose fluids containing particles or beads. Applications like colour printing with emulsion paint or medical applications containing beads for medical detection require devices particularly insensitive against clogging.

System Description

Key element of the dispensing device termed "PipeJet™" is a clamped commercially available polymer tube. The rear end of the tube is connected to a reservoir and serves as a supply channel. The front end forms the nozzle for liquid ejection. By a fast displacement of a piston connected to a piezostack actuator, the tube is squeezed on an active area of approximately 5 mm length displacing the liquid towards both ends of the tube (figure 1). Due to the low fluidic resistance between the actuation area and the nozzle, most of the displaced volume is ejected as free liquid jet like shown in figure 2. By varying the dimension or the position of the active area compared to the nozzle outlet a

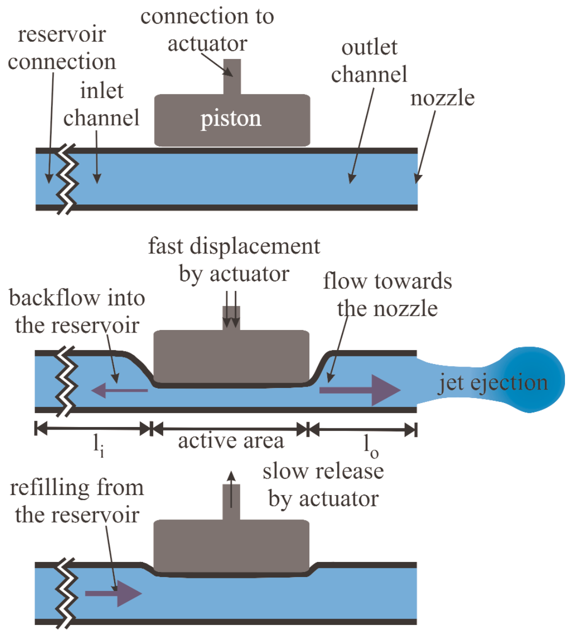

Fig. 1: Working principle of the PipeJet™ dispenser.

variation in the ejected droplet volume can be achieved. A chamfered active area of the actuator towards the nozzle outlet additionally leads to a flow in this preferred direction and therefore a higher dosage volume. After releasing the squeezing of the tube the refill is done by capillary forces only.

Results

The working principle has been proven by stroboscopic and gravimetric measurements as displayed in figure 2 and 3. For the experiments we used a tube with 200 µm inner diameter and wall thickness of 25 µm. At a piezostack displacement of 40 µm and a displacement velocity of 90 µm/ms droplets of approximately 22.5 nl have been obtained. Obviously the droplet ejection is very stable, no tail or satellite droplets can be observed. Recording a stroboscopic photo sequence leads to a fluent video that indicates that every ejected droplet is very similar in shape and velocity. The standard deviation of the dosage volume of 1.6% is excellent (cf. figure 3). As confirmed in a long term

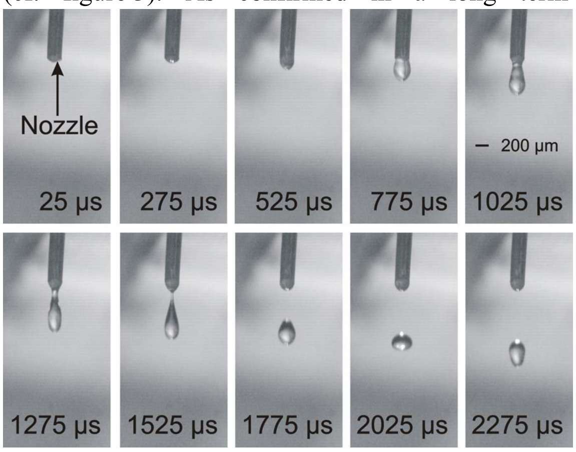

Fig. 2: *Stroboscopic pictures of an ejection process.*

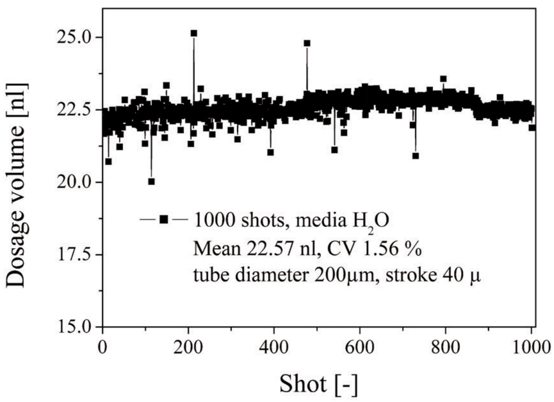

Fig. 3: *Long term measurement of the dosage volume at a frequency of 0.1 Hz.*

experiment the performance stays stable over at least 1,000,000 cycles.

One reason for the good performance is that there is nearly no material surrounding the nozzle like in other dispenser devices [1,2] except the thickness of the tube. Furthermore the tube has a "perfect" circular shape with no edges leading to an optimum flow field.

Different media in the viscosity range from 1 mPas to 27 mPas have been dispensed successfully as can be seen in figure 4. We achieved a dosage volume nearly independent on viscosity up to 20 mPas similar to [1]. The dispenser even worked with methanol, which is usually difficult to handle due to its low surface tension. The frequency characteristics shows an frequency independent dosage volume up to frequencies of 15 Hz (figure 5) with a 200 µm tube. At higher frequencies it is assumed that due to the significant flow resistance of the tube capillary refill limits the ejected volume.

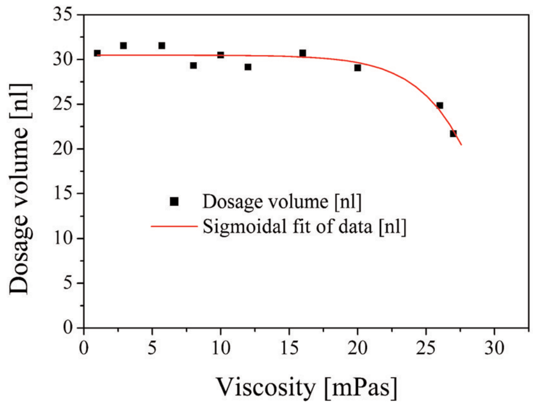

Fig. 4: *Dosage volume depending on the viscosity at a frequency of 0.1 Hz with 50 µm stroke and different glycerol/water dilutions getting the different viscosities.*

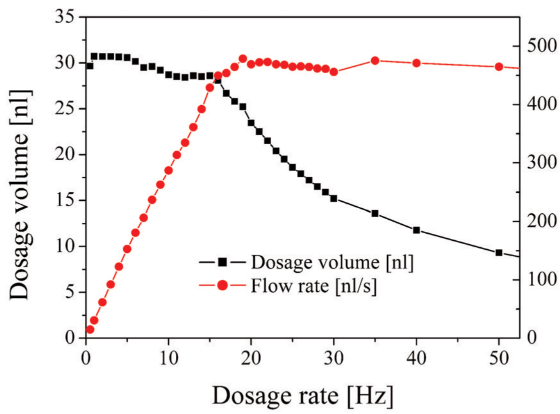

Fig. 5: *Frequency dependency of the dosage volume of H_20 at a piezo stroke of 50 µm.*

With a more powerful actuation electronics and actuator and therefore a higher and faster piezostack stroke it was even possible to dispense a motor oil with a dynamic viscosity of about 72 mPas as well as an emulsion paint with a particle size of approximately 40 µm and a dynamic viscosity of approximately 200 mPas at a frequency of 1 Hz. Surprisingly but fortunately no clogging was observed using a 200 µm tube.

The above mentioned electronics for actuating the piezostack provide a maximum charging current of 8 A with a minimum time step size of 1 µs and a very short relaxation time of only 400 ns. With this performance the piezostack can be operated at a maximum cycle frequency of about 340 Hz with a maximum stroke of 160 µm and velocity of 500 µm/ms.

The ejected droplet volume can be defined varying different parameters like the tube diameter, the piston stroke, the dimension and the shape of the active area and finally the fraction of inlet and nozzle channel length. Increasing the piston stroke, the active area or the fraction of inlet and nozzle channel length l_i/l_o leads to an increase of the ejected volume. In figure 6 the dependency of the dosage volume on the piston stroke is displayed. Obviously there is a linear correlation between the dosage volume and the piezostack stroke. The comparison with the simulation results which are also illustrated in figure 6 leads to a good agreement in the stroke characteristics.

To increase the flow rate of the PipeJet™ shorter tubes with larger inner diameter have been used. Using a 500 µm tube, a dosage volume of about 53 nl was obtained. With this setup flow rates up to 15,7 µl/s have been reached for operating frequencies up to 290 Hz. For a tube of 1000 µm diameter, the maximum obtainable flow rate was found to be 143 µl/s at a droplet volume of 420 nl with a frequency of 340 Hz.

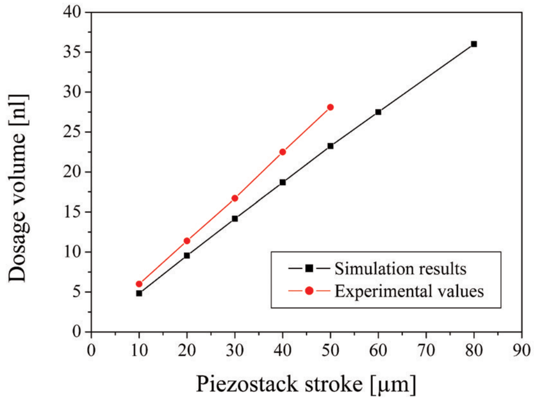

Fig. 6: *Measured and simulated dosage volume for H_2O depending on the piezostack stroke at a frequency of 0.1 Hz. (l_i/l_o = 8 for the simulations.)*

Simulations

Computational Fluid Dynamic (CFD) simulations of the dispensing process have been performed using the movement of the stroke as boundary condition like it is displayed in figure 7. This way it has been proven that the dosage volume depends linearly on the actuator stroke resulting in a variability of dosage volume from 4.8 nl to 36 nl (cf. figure 6). Simulations worked well up to a stroke of 80 µm which corresponds to 40% of the tube diameter. A larger displacement was not possible because of convergence problems of the simulation algorithm.

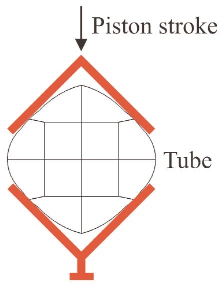

Fig. 7: *Boundary condition for the simulations and resulting squeezing of the tube.*

In order to find an optimum of the tubing dimensions we simulated different ratios of inlet length l_i to outlet length l_o (figure 8). We found that the dosage volume can be varied from 13 nl ($l_i/l_o = 1$) to 19 nl ($l_i/l_o = 8$) at a stroke of 40 µm.

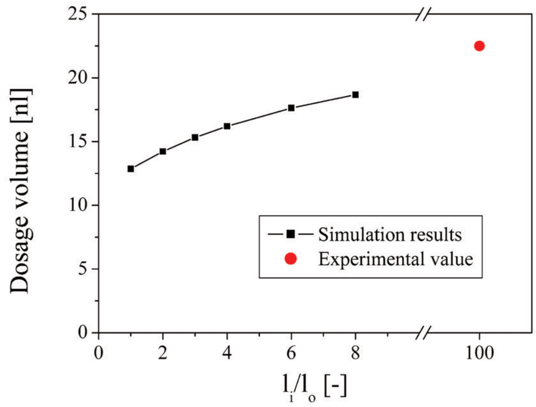

Fig. 8: *Simulated dosage volume of H_2O depending on l_i/l_o, where l_i and l_o are the length of the inlet and outlet channel respectively compared to the experimental value (40 µm stroke).*

Comparing the simulated volume of 19 nl ($l_i/l_o = 8$) with the gravimetric measured droplet volume of 22.5 nl with the same piezo stroke leads to a deviation of 15%. This disagreement is supposed to be caused by the fact that the fraction of inlet and outlet channel in the real experimental setup was considerably higher (cf. figure 8). For the experimental value the l_i/l_o fraction is estimated to be 100.

To have also a qualitative comparison of experimental droplet shapes and simulation further simulations were performed using the Volume of Fluid (VOF) method [6]. In this case it was necessary to substitute the moving boundary condition with an equivalent pressure boundary condition to avoid convergence problems caused by the fluid-structure interaction (FSI). Therefore the pressure inside the tube at the moving grid boundary conditions (cf. figure 7) were recorded. This pressure characteristics were then used as boundary condition for the new simulation without accounting for the tube deformation anymore. Comparing to the stroboscopic pictures (cf. figure 2) the shape of the ejected droplet depicted in figure 9 shows a quite good agreement. Unfortunately there are some discrepancies in the time behaviour. The simulation indicates a faster droplet ejection compared to the experiments. The reason for this might be the applied pressure boundary condition. Further investigations are required to clarify this point and to improve the CFD model.

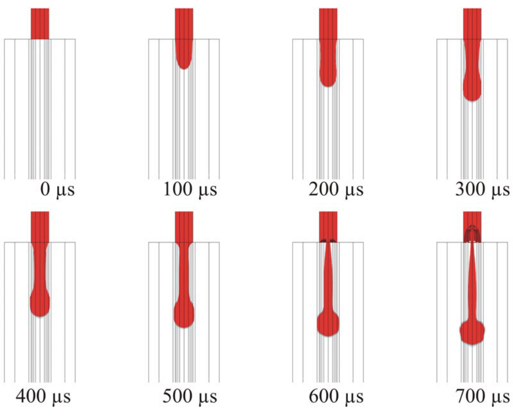

Fig. 9: *Simulation results of a droplet ejection of* H_20 *using the Volume of Fluid (VOF) method* ($l_i/l_o = 8$, *40 µm stroke).*

Conclusions

For the first time we presented a new nanoliter dispenser based on a low-cost polymer tube. Compared to all known micromachined dispensers the PipeJet™ is cheap and simple. A wide dynamic viscosity range from 1 mPas to 200 mPas has been printed successfully so far. Furthermore the dosage volume is nearly independent of the viscosity up to 20 mPas. The standard deviation of less than 2% within 1,000 shots is excellent. The presented device provides also the possibility to dispense emulsions containing particles or beads up to 1/5 of the tube diameter without clogging.

Possible applications for the PipeJet™ are all areas where the handling of high viscous fluids is necessary like some medical or biological applications, oil, adhesive or liquid metal dosing. Moreover all kinds of particle or bead dispensing like magnetic beads for life-science applications or emulsion paint can be achieved. Apart from its extraordinary performance in the nanoliter range the ultimate advantage of the technology is the low-cost fluidic part formed by the tube. This makes the device especially suited for disposable applications.

Acknowledgement

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