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A microscale pulsatile flow device for dynamic cross-slot rheometry

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ABSTRACT

The design of micropumps received full attention since micromanufacturing and microfluidics techniques have become part of the engineering toolbox. The focus of most studies has been on the efficiency of these pumps: maximal net (mean) flow at minimal input power.

We introduce a pulsatile micropump system, that is designed to dynamically perfuse a cross-slot microrheometer. To characterize complex materials dynamically, unsteady (oscillating and pulsating) flows with a frequency range of 0.1–20 Hz at amplitudes of 10–100 nl/s are required. Hence, in our study the priority concerning micropumps shifts from efficiency to the ability to produce well-defined flow pulses.

For this purpose, an oscillatory micropump, based on a deflecting diaphragm, is designed and tested. By periodically deflecting a steel plate into a rigid fluidic chamber using a voice coil, an oscillatory flow is produced. Plate deflection is governed by bending, such that the stroke volume is proportional to the current through the voice coil. The oscillatory flow is superimposed to the steady flow of a syringe pump. The pump system obtained is characterized by micro particle image velocimetry (μ-PIV) measurements using fluorescence microscopy.

The results show that the superposition of the mean flow of the syringe pump and an oscillatory flow of the diaphragm pump, is valid. A linear scaling of flow amplitude with frequency and driving voltage is found for frequencies up to 4 Hz, after which excessive damping takes place. The causes for this behavior are identified and explain the results well. With this information, amplitude scaling for sinusoidal flow waves of different frequencies can be performed. In conclusion, with the present system, pulsatile flow with a well-defined waveform and a dynamic range up to 16 Hz, can be created in an open-loop driven fashion.

1. Introduction

The last 30 years, various micromachining techniques were introduced, which opened the field of MEMS and microfluidics. The need to displace fluids at small scale followed, which resulted in the design and evaluation of dozens of microscale pumps (for extensive reviews, see [1–7]). In *displacement* micropumps, a moving boundary or liquid displacement exerts a pressure force, while *dynamic* micropumps rely on a direct energy transfer to the fluid to be pumped. The former usually generate pulsatile flow, the latter drive the fluid in a continuous, constant manner [3]. The focus here is on diaphragm displacement micropumps, where a solid diaphragm is deflected into a valve chamber to push fluid forward. These pumps can be equipped with passive check valves with moving parts (flaps [8] or balls [9,10]), or non-moving valves (e.g. Tesla microvalve [11]), or be valveless (nozzle-diffuser design [12,13], vortex areas [14]).

Actuation is performed in several ways, among which piezoelectric [15–18] and electromagnetic or voice coil actuators [9,12,15,19] are most popular. Often, the pump is integrated with a fluidic system, ranging from lab-on-a-chip [20] or micro-total-analysis systems (μ-FACS, -ELISA or -mass-spectrometer [21]) to miniaturized fossil fuel cell batteries [22].

During characterization of these pumps, the focus has been on the efficiency and absolute net flow: how can a maximal net flow be produced with a minimal amount of power input. However, for some applications, such as the culturing of endothelial cells in a microfluidic channel, the periodic pulsatility is of interest, whereas efficiency is of secondary importance. Another application

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symbol	description, unit
A	transfer function gain, –
a	diaphragm radius, m
a_t	connecting tube radius, m
C	compliance, $m^4\,s^2\,kg^{-1}$
C_{sat}	compl. saturated fluid, $m^4\,s^2\,kg^{-1}$
D	plate constant, m
D_H	hydraulic diameter, m
E	Young's modulus, Pa
F_L	Lorentz force, N
f	flow frequency, Hz
f_0	natural frequency, Hz
f_c	cutoff frequency, Hz
f_{res}	resonance frequency, Hz
H	channel height, m
H_c	height pump chamber, m
h	diaphragm thickness, m
I	inertance, $kg\,m^{-4}$
I_c	inertance, $kg\,m^{-4}$
k	diaphr. spring constant, $N\,m^{-1}$
L	tube length, m
L_t	tube length, m
M	equivalent mass, kg
$p_{1,2}$	transfer function poles, –
$\bar{Q}$	mean flow, $m^3\,s^{-1}$
$Q_{1,2}$	outflow of cross-slot, $m^3\,s^{-1}$
Q_{out}	systemic outflow, $m^3\,s^{-1}$
Q_{pulse}	pulsatile flow, $m^3\,s^{-1}$
$Q_{syringe}$	syringe pump flow, $m^3\,s^{-1}$
$R_{(1,2)}$	hydraulic resistance, $kg\,m^{-4}\,s^{-1}$
Re_{max}	(max.) Reynolds number, –
r	radial coordinate diaphr., m
r_0	piston radius, m
s	Laplace parameter, –
V_{max}	maximum velocity, ms^{-1}
W	channel width, m
$y(r)$	diaphragm deflection, m
α	Womersley number, –
ΔV_{stroke}	stroke volume, m^3
ζ	damping constant, –
ν	kinematic viscosity, $m^2\,s^{-1}$
ν_p	Poison ratio, –
ρ	fluid density, $kg\,m^3$
ω	angular velocity, $rad\,s^{-1}$
ω_n	natural frequency, $rad\,s^{-1}$

is experimental characterization of dispersion-induced boundary layer mixing, demanding two well-controlled pulsatile flows in counter-phase [23].

In the current work, a device is developed that can serve as a pulsatile flow pump for a microfluidic rheometer, in which complex micromaterial behavior (microgels, cells, elastic capsules, droplets) can be mechanically probed. The micro-rheometer is based on a cross-slot setup (see [24,25] for an application with visco-elastic drops and diluted polymer solutions, respectively), which consists of a microfluidic chip with crossing channels, a metering valve, and a feedback system. With the cross-slot setup, in which elongational flow exists, such as in the four roll mill [26], a particle can be captured (Fig. 1, similar to the hydrodynamic trap of [27]). When captured, hydrodynamic forces are used to apply stress to the particle. From the observation of the resulting deformation of the particle, its rheological properties can be determined.

In this work, the device is designed to be applied to the dynamical probing of red blood cell mechanics. To extend the cross-slot principle towards a micro-rheometry setup, the inflow of the device should be made pulsatile (non-zero mean flow) with well-controlled period and amplitude, such that frequency-dependent behavior of the red blood cell under investigation can be characterized. Concerning the red blood cell membrane, which has (relaxation) time constants of 0.15–0.5 s, a pump that drives pulsatile flows with a frequency of about 20 Hz, and amplitudes down to 10 nl/s, is required. It is also essential that the waveform of the flow is purely sinusoidal, such that the cell can be probed at a single frequency. Although literature concerning micropumps is abundant, to our knowledge no data of instantaneous flows (i.e. waveforms) at these scales have been published.

State-of-the art syringe pumps could not deliver sinusoidal pulsatile flows via the programming function (Harvard PHD 2000, Harvard Apparatus) or the command line option (Nexus 3000, Chemyx). Hence, a pulsatile micropump system is needed, with which the flow waveform is well-controlled and easily tunable. To our knowledge, such a device is neither commercially available, nor reported about in literature.

Altogether, our goal is to design a pump system, which produces pulsatile flows that are well-controlled in terms of amplitude, frequency, and pulse shape. In the rest of this paper, a motivation for the specific design is given, in which the flow is produced in an open-loop driven way. During the design phase, aspects concerning geometry, system dynamics, and fluid actuation, are taken into account.

Next, to assess the quality of the waveforms the microscale flows are measured with high temporal resolution, using micro particle image velocimetry (μ-PIV). Third, the results of steady, oscillatory, and pulsatile flow are presented, as well as passive and active noise responses. This should demonstrate the behavior of the pump system, as well as its capabilities. The dynamic behavior of the system can be explained by discussing different physical phenomena. These insights can be used to gain full control over the flow waveform.

2. Materials and methods

2.1. Pump system design

The exact computation of impedance in oscillatory microflow is not straightforward [28]. However, we assumed that our system is linear and time invariant (LTI system), such that a pulsatile flow can be generated using a syringe pump and a reciprocal diaphragm pump in series: according to the superposition principle the constant and oscillating flow are summed [29]. This has successfully been applied in larger linear hydraulic systems [30].

The produced pulsatile flow of the pump system equals the stroke volume per unit of time: $Q_{pulse} = Q_{syringe} + \lim_{\Delta t \to 0} \Delta V_{stroke}/\Delta t$. The required flows impose restrictions to the geometry and dimensions of the pump (Fig. 2a). The maximum flow amplitude is dependent on the pulse frequency. When the mean flow $\bar{Q}$ = 100 nl/s, and the minimum in the pulse lies at zero, the necessary stroke volume $V_{stroke} = \bar{Q} \cdot 1/2\pi f$. This implies that a frequency of 0.1 Hz demands a volume displacement of 159 nl. On the other hand, at higher frequencies and low amplitudes, the necessary stroke volume is reduced and noise suppression becomes more important.

To be in the right range of frequencies and flows, an axisymmetric valve chamber (radius a = 6.5 mm, height H_c = 2 mm) holds a circular plate with a thickness of h = 200 μm. An O-ring between the diaphragm and the lid of the pump, which is fully compressed to make steel-to-steel contract, ensures sealing of the chamber.