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LCAT pump optimization for an integrated microfluidic droplet generator

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Abstract

We demonstrate an on-chip integrated droplet generator enabled by lateral cavity acoustic transducer (LCAT) oil and water microfluidic pumps. Both oil-in-water (O/W) and water-in-oil (W/O) droplet generation are demonstrated. The LCAT pumps are energized by piezoelectric acoustic energy to induce rectified microstreaming for pumping liquid. In this work, the analysis of geometric optimization of LCAT pumps was performed. The LCAT droplet generator was characterized in terms of size and frequency of generated droplets. For the W/O droplet generation, the controllable range of droplet diameter was 50–420 μm ; while for the O/W droplet generation, the controllable range of droplet diameter is 60–150 μm . The minimum voltage for stable droplet generation can be as low as 4 V_{pp} . The first LCAT pump for pumping oil is also demonstrated by lipophilic treatment of the microfluidic channel. The integrated LCAT droplet generator offers a valveless, portable, low-cost, and low-power platform for generating microfluidic droplets. The LCAT droplet generator can be a key enabling microfluidic component towards the realization of a portable diagnostic/screening platform.

Introduction

Over the past decade, microfluidic droplets have attracted tremendous attention and have been practically applied to medical diagnoses, biochemical assays, material syntheses, and manufacturing in various fields, as well as to multidisciplinary scientific studies. This is owing to their remarkable traits such as ultra-tiny volumes (nL–fL), large surface-area-to-volume ratio, great monodispersity, compartmentalization, and interfacial modification (Teh et al. 2008; Theberge et al. 2010; Duncanson et al. 2012; Lagus and Edd 2013). For biochemical/medical applications, microfluidic droplets can be produced as storages (e.g. a droplet library) to contain various reagents or samples (Trivedi et al. 2010), as microreactors (for PCR detection (Hatch et al. 2011), monitoring of protein crystallization (Li and Ismagilov 2010) or enhancement of DNA hybridization and detection (Fang et al. 2012)), as a micro-compartment for single cell analysis or drug screening (Joensson and Svahn 2012), or as an artificial cell for forming a biomimetic environment (Zhang et al. 2013). Most of these microfluidic droplet devices lack on-chip pumps and mainly rely on bulky and energy-consuming syringe pumps, compressed-air (pneumatic) systems, or centrifugal systems to generate droplets. This presents a major limitation in introducing chip-scale droplet-based

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systems to a broader market such as point-of-care testing (POCT) or portable in-vitro diagnostics (IVD).

A few devices that possess the potential for portability have been proposed to generate O/W (oil-in-water) or W/O (water-in-oil) droplets, which are powered by electric field (e.g., electrowetting on dielectric, EWOD) (Choi et al. 2012), piezoelectric actuator (Bransky et al. 2009), dielectrophoresis (Schwartz et al. 2004) or finger power (Iwai et al. 2011). For both the EWOD device and the piezoelectric actuator device, their chip fabrication is sophisticated and costly, and relatively high applied voltage (usually larger than 40 V) is required to generate droplets (Brassard et al. 2008). Although the finger-power droplet generator is simple to operate manually, the size and generation rate of resulting droplets cannot be precisely controlled. Our previous work realized a powerless pump for droplet generation (Bilotkach and Lee et al. 2008). It was a simple device but unable to precisely control the droplet generation. Currently, none of the portable devices that produce O/W (or W/O) droplets offer low cost, low power consumption, as well as accurate controllability in a simple device. Since recently there has been a growing tendency to harness portable microfluidic devices for a variety of applications (Chin et al. 2012), development of a portable droplet generator is critical to many of these applications.

Acoustic-induced microstreaming flow surrounding a liquid-air (bubble) interface was effectively utilized to enhance DNA hybridization (Liu et al. 2003), lyse cells (Marmottant and Hilgenfeldt 2003), and sort particles (Wang et al. 2011). Based on this mechanism of microstreaming flow, we developed the lateral-cavity-acoustic-transducer (LCAT) technique (Tovar and Lee 2009), and exploited it to pump aqueous liquids (Tovar et al. 2011), shorten the incubation period of protein assay (Lee et al. 2010), extract plasma from whole blood (Doria et al. 2011), switch particles (Patel et al. 2012), and shear DNA (Okabe and Lee 2013) in simple microfluidic devices. Samples within the LCAT chip can be manipulated via the microstreaming flow induced by acoustic waves from piezoelectric transducers (PZTs) that are located beneath the chip and energized through a battery or USB powered circuit (Tovar et al. 2011). Additionally, the design and fabrication (one-step photolithography process) of LCAT chips are so simple that it can be integrated with existing microfluidic devices. The LCAT technology is therefore a promising means to realize a hand-held diagnostic system or a compact scientific instrument.

Herein, we invented a novel microfluidic device entitled the LCAT droplet generator for the generation of O/W or W/O droplets. For the W/O droplet generation, droplets can be controlled to the range of 50–420 μm in diameter (65 pL to 39 nL in volume) while for the O/W droplet generation, droplets range from 60–150 μm in diameter (113 pL to 1.77 nL in volume). The voltage to generate droplet can be as low as 4 V_{pp} . The LCAT droplet generator offers low-cost fabrication, portability, and a valveless on-chip pumping scheme. We believe the LCAT droplet generator can be facilely applied to or integrated with other components or systems to facilitate the realization of a portable diagnostic/screening platform. Geometric effects on the performance of LCAT pump, as well as lipophobic surface treatment of the LCAT pump for pumping oil are also investigated in this work.

Method

Device design and experiments

Figure 1 illustrates an LCAT droplet generator composed of two LCAT pumps for pumping aqueous and oil liquids into a T-shaped junction for droplet generation. The LCAT pump consists of a main channel 86 mm in length and 96 pairs of dead-end side channels (or lateral cavities). Once liquid is primed over through the main channel, the liquid-gas interface spontaneously forms at the junctions of the main and side channels. When an acoustic wave generated from a PZT transfers to the interface, the interface consequently vibrates, followed by a first-order pulsatile flow within liquid in close vicinity to the interface. This first-order pulsatile flow thereupon brings along a second-order microstreaming flow due to the force balance between the nonlinear inertia term and the viscous term of liquid (Marmottant and Hilgenfeldt 2003). Due to the designed angle between the side and main channels, this microstreaming flow is rectified yielding a positive pressure difference between upstream and downstream, resulting in a downstream net bulk flow. The droplet generation region is designed to include buffer chambers for dispersing perturbations in flow near the T-shaped channel with a constriction ($25\text{ }\mu\text{m} \times 25\text{ }\mu\text{m}$) that facilitates droplet generation.

The LCAT droplet generator is energized by two PZTs coated with a $\sim 300\text{ }\mu\text{m}$ thick layer of ultrasound gel (Parker Laboratories, Inc., USA). The cuboidal PZTs (resonant frequency of 67.7 kHz) were manufactured by trimming cylindrical PZTs (SMD50T30F45S, Steiner and Martins, Inc., USA) with a handheld tool (Dremel 4000, Dremel Company, USA), and mounted on a homemade acrylic stage. Two function generators (Agilent 33220A, Agilent Technologies, Inc., USA) output square-wave signal at 67.7 kHz to separately control the two PZTs which generate acoustic energy to drive two LCAT pumps. No coupling effect of acoustic waves from the separate PZTs was observed when the PZTs were in operation. For the generation of W/O droplets, we used DI water as the dispersed phase and mineral oil ($\rho=0.838\text{ g/mL}$ and $\mu \sim 20\text{ cp}$ at $25\text{ }^{\circ}\text{C}$; Mineral oil-light, Sigma-Aldrich Co. LLC.) with 2% Span 80 (Sigma-Aldrich Co. LLC.) as the continuous phase, and for the O/W droplets, we use mineral oil as the dispersed phase and DI water with 2% Tween 20 (Sigma-Aldrich Co. LLC.) as the continuous phase. The droplet generation and interfacial vibration were individually visualized using an upright microscope (Nikon Eclipse L150, Japan) equipped with a $5\times$ objective (CFI LU Plan Fluor, NA = 0.15) and a $10\times$ objective (CFI Plan Fluor, NA = 0.3). A high-speed camera (Phantom 7.3, Vision Research, Inc., USA) was used to capture images.

Our previous work successfully applied the LCAT techniques to various applications, but barely focused on optimization of LCAT devices. Previous studies reveal that a droplet flow must bear a larger resistance (pressure drop) than a single-phase continuous flow over the same distance (Adzima and Velankar 2006). Therefore, in this work, we explore the geometric effects on pumping performance of the LCAT pump to procure a relative optimized design for an LCAT droplet generator. Figure 2 shows five LCAT pumps with different side channel designs, specifically named device 1 (D1), device 2 (D2), device 3 (D3), device4 (D4), and device 5 (D5). For all these devices, the depth of the channel was

100 μm , the width of the main channel was 500 μm , and the angle between the main channel and the side channel was 15°(optimized angle (Tovar and Lee 2009)). The width, height, and length of the main channel determine the flow resistance, which in turn affect the performance of the LCAT pump. The length of channel is proportional to the resistance while the width and height of channel are inversely proportional to the resistance. We fabricated the channel with a height of 100 μm , which is typical in microfluidic devices. We chose 500 μm as the width of the channel for a reduced flow resistance in the channel resulting in higher pumping flow rates. As for the geometric analysis, parameters encompass the width (a) of the junction between main channel and side channel, the spacing (b) between two side channels, and the length (c) of side channel.

Fabrication

The LCAT droplet generator is fabricated using the standard soft lithography procedure, which is composed of polydimethylsiloxane (PDMS) as the top layer and a 150 μm thick cover glass as the bottom layer. The PDMS layer with the microchannels are molded from an SU-8 master fabricated on a silicon wafer. SU-8 2050 (Microchem Corp., USA) is spun coated according to manufacturer's data sheet to obtain a photoresist layer thickness of 100 μm on the wafer. This is exposed using UV radiation through a transparency mask to produce the desired microchannel patterns before its development using SU-8 developer (Microchem Corp., USA). A 10 : 1 ratio of PDMS base to curing agent (Sylgard 184, Dow Corning Corp., USA) is mixed, degassed and poured onto the SU-8 master and cured in a 80°C oven for 4 h. After the solidification, the PDMS layer is peeled from the master. The PDMS layer is trimmed, holed and plasma bonded (2 min air plasma treatment) to a cover glass. The device is then placed in a 120°C oven overnight before being used for experimentation.

Results and discussion

Performance (pQ curve or head-capacity curve) of LCAT pumps

In order to fully comprehend the performance of a pump, the pump must be evaluated against a varied load/resistance so as to obtain the correlation between the differential pressure across the inlet and outlet (total head, $P = P_{in} - P_{out}$) and volumetric flow rate (Q), called the PQ curve. Figure 3 depicts the measured performance of the LCAT pump. The pressure (P) delivered by the pump is mainly employed to pump liquid to flow through channel and tubing under a hydraulic load. The flow channel pressure drop P_d can be characterized by the volumetric flow rate (Q) and a total resistance (R), so that $P_d = QR$. The hydraulic load can be regarded as the gravitational pressure defined as $P_g = \rho gH$, where ρ is the liquid density (water, 998 kg/m^3 at room temperature), g is the local gravity (9.8 m/s^2), and H is the differential height between inlet and outlet water levels. During LCAT pump operation, the differential pressure (P) over time is given by

$$P(t) = \Delta P_d(t) + \Delta P_g(t) = Q(t)R + \rho gH(t) = AH'(t)R + \rho gH(t), \quad (1)$$

where A is the inner cross-section area of tubing ($\sim 1.8 \times 10^{-3} \text{ mm}^2$), and $H'(t)$ is the differentiation of $H(t)$ indicating the rising velocity of water level along the tubing.

The cross-section of channel is assumed to be a rectangle with the aspect ratio (α) of height (h) to width (w), 0.2, and the tubing has a round inner cross-section. On the basis of pressure driven and laminar flow (Reynolds number $Re < 1$), the R can be derived as shown (Dentry et al. 2014).

$$R = R_1 + R_2 \approx \frac{12\mu L\alpha}{h^4} \frac{1}{(1 - 0.63\alpha)} + \frac{128\mu H(t)}{\pi d^4}, \quad (2)$$

where R_1 is the flow resistance of fluid in the channel while R_2 is that in the tubing, μ is the dynamic viscosity of liquid (water, 0.000855 kg/m-s at room temperature), L is the length of main channel (86 mm), and d is the diameter of tubing (480 μm).

The $H(t)$ was measured when the LCAT pump operates with an applied voltage 3.0 V_{pp} over time (t from 0 to 360 s, sampling rate is 30 s per point) and then calculate P based on equation 1. Figure 4 shows the correlation between P and Q for various LCAT pumps. We performed 5 independent measurements for each test, and the ranges of error for these tests were below 20%. It is observed that for all pumps, the total pressure drop P is linear to Q which can be described by $P = mQ + C$, where m is the slope of a PQ curve and C represents the maximum back pressure P_{max} that a LCAT pump can generate. This linear correlation may be an innate characteristic of an acoustofluidic pump (Dentry et al. 2014), which implies that the flow inside the LCAT pump is not altered drastically by a slight increase in the load resistance. For the evaluation of a hydraulic pump, the maximum back pressure P_{max} , the maximum volumetric flow rate Q_{max} , and the dynamic range of volumetric flow rate that a pump yields should be taken into account. The larger value of Q_{max} and P_{max} , the wider dynamic range the pump can generate (Laser and Santiago 2004), which defines the performance of the LCAT pump. Comparing the PQ curves of all devices, it is evident that D4 possesses the largest Q_{max} and P_{max} while D5 possesses the smallest Q_{max} and P_{max} ; the values of Q_{max} and P_{max} for other devices are between those for D4 and D5. Furthermore, D4 has the smallest absolute value of m , which indicates its performance curve has the widest dynamic range of flow for a given pressure range. It is therefore concluded that the order of pump performance is $D4 > D3 > D2 > D1 > D5$ according to the comparison of PQ curves in figure 4.

Interfacial vibration associated to performance of LCAT pumps

To understand the underlying mechanism that affects the performance of LCAT pumps, we looked into how the behavior of acoustic-induced interfacial vibration correlated with the performance of the pumps. Marmottant and Hilgenfeldt (2004) reported the experimental and theoretical oscillating behavior of a bubble placed in an isolated pit. The typical velocity scale of a microstreaming flow nearby a boundary of oscillating bubble is given by

$$u_s \sim \varepsilon^2 r \omega = A r \omega, \quad (3)$$

where ε is a proportional constant pertinent to the oscillation amplitude $A (= \varepsilon r)$, r is the radius of a bubble, and ω is the angular frequency. Figure 5 shows interfacial width (w) and oscillation amplitudes (A) of liquid-air interface for various LCAT pumps. For the LCAT pump, the flow rate of a bulk flow (Q) is assumed to be positively correlated to u_s , and based on equation 3, we derive the relationship $u_s \propto A r \omega$. Q is therefore assumed to be positively correlated to A , w , and ω .

The acoustic frequencies for all tests were fixed at 67.7 kHz, meaning ω is constant. Q_{max} (at $t = 0$ s) of D1, D2, D3, D4, and D5 was calculated as 8.8, 9.6, 10.4, 24.5, and 3.6 $\mu\text{L}/\text{min}$, respectively. The pumps D1, D2, and D3 are with similar geometric designs except that the ratio of a (junction width) to b (spacing between two side channels) as shown in figure 4. The pumps were with similar oscillating amplitudes (13–14 μm), but the interfacial widths were different ($w_3 > w_2 > w_1$, the larger a leads to the larger w) due to different a/b ratios. It was found that Q_{max} increased with w , revealing that Q_{max} is positively correlated to w . This was due to a larger interface (w) allowing more acoustic energy transferred to create microstreaming flow which, in turn, induces a stronger bulk flow with a larger flow rate. Another reason is that the larger liquid-air interfacial area renders more slip boundary area, leading to reduction of the frictional drag as liquid flows through the channel (Karatay et al. 2013). The pumps D3, D4, and D5 are with different side channel length c but same width a which led to all pumps having similar interfacial widths and different oscillating amplitudes ($A_4 > A_3 > A_5$); this difference in oscillating amplitudes reveals that c is negatively correlated to A . Also, it was found that Q_{max} increased with the increasing of A . c is therefore negatively correlated to Q_{max} . It is not clearly evident why there is this negative correlation between c and Q_{max} . We infer that the larger c value might incur more acoustic energy dissipation, thus reducing the microstreaming flow and the bulk flow rate.

From the aforementioned analysis, we hypothesized that the performance of LCAT pump could be enhanced by increasing the width of junction (a) or by decreasing the length of side channel (c). Compared to other pumps, D4 had the largest a and shortest c which posed a superposition effect on the improvement of performance, which was much superior to those of other pumps (see figure 4). We hence designed a LCAT droplet generator based on the dimensions of pump D4.

Surface treatment with FDTS vapor deposition

To date, according to our knowledge, there have been no on-chip integrated micro pumps to pump oil reported in the literature. As oil is innately viscous, leading to a large resistance against the driving force, pumping oil liquids with an on-chip pump is challenging. In this work, we demonstrate pumping oil liquids with our LCAT pumps. In addition to the high viscosity, the affinity of oil liquids at different interfaces is a critical variable. Most oil liquids favor PDMS or glass substrate and form contact angles smaller than 90° due to a strong attractive force between liquid and solid molecules. Since the LCAT technique

utilizes the interfacial vibration to induce liquid flow, the formation of liquid-air interface is pivotal.

We performed a set of repetitive tests ($N=5$) for the contact angle measurement. For a water drop, its contact angle (θ) on the PDMS surface without any treatment is $104\pm1^\circ$ (hydrophobic) while that for a mineral oil drop is $48\pm1^\circ$ (lipophilic) as shown in figure 6. If mineral oil is primed in channels of the LCAT pump without surface treatment, the oil is prone to flowing into side channels resulting in receding liquid-gas interfaces and the forming defective air pockets. Without complete air pockets, it is impossible to generate effective microstreaming flow to pump liquid.

For successful pumping of oils, it is imperative to surface treat the side channels of devices to reduce oil wetting. We adopt the vapor deposition procedure to treat channels of LCAT pump with perfluorodecyltrichlorosilane (FDTS, $C_{10}H_4Cl_3F_{17}Si$, SIH5841.0, Gelest Inc.) (Ghosh et al. 2009). After plasma bonding the PDMS device to glass slides, 1 μ L FDTS is dripped into the inlet reservoir of LCAT device and then the reservoir is sealed with tape. Then the device is placed into a low-pressure desiccator at 65°C for overnight treatment. The lifetime of the FDTS-treated surface is nominally a couple of days. All tests were executed within 24 hours after the surface treatment.

The surface coated with FDTS monolayer possesses much lower surface energy for most liquids such as water, mineral oil, oleic acid, triacetin and so on. Therefore, treated surfaces have a greater repellent property for both water and oil. A water drop on a FDTS treatment surface has $\theta = 107\pm2^\circ$ while a mineral oil drop on that surface has $\theta = 97\pm3^\circ$ (see figure 6), which demonstrates that both liquids are less wetting to the FDTS treated surface. Compared to the channel without the FDTS treatment, the FDTS treated channel enables the formation of perfect air pockets after being filled with mineral oil, and the pockets are able to retain air-liquid interface for longer than 30 min. By means of the FDTS treatment, the LCAT pump for oil liquids is made possible. The oil flow in the FDTS-treated channel was quite stable.

Performance analysis of the LCAT pump D4 for pumping aqueous and oil liquids under various applied voltages

Since the LCAT droplet generator is based on the D4 design, its performance for pumping aqueous and oil liquids under various applied voltages was investigated as shown in figure 7. The maximum back pressure, P_{max} was achieved when Q is 0. For pumping aqueous or oily fluids with the pump D4, P_{max} was found to increase linearly with applied voltage (both R-squared values are 0.97). Increasing the applied voltage on the pump increased the interfacial amplitude (A) which induced a larger back pressure in the device. If the applied voltage to pump water (V_{water}) exceeded $6.0 V_{pp}$, the liquid-air interfaces (air pockets) would become unstable due to the interfacial tension unable to bear the strong oscillation and to retain the integrity of interfaces. The air pockets consequently broke or expanded beyond the side channels into the main channel. Owing to the large viscosity (~ 20 cp) of mineral oil, the pump needed a much larger applied voltage to pump the oil to reach a certain level of back pressure, compared to pumping water (viscosity ~ 1 cp). If the applied voltage for oil (V_{oil}) was smaller than $2.0 V_{pp}$, the driving force, generated from the pump,

was too small to overcome the viscous and frictional forces, and thus no back pressure could be measured. Likewise, once the voltage applied to the pump for pumping mineral oil was over a threshold of $16.0 V_{pp}$, the flow system becomes unstable.

W/O droplet generation

Figure 8 shows the W/O droplet generation with the LCAT droplet generator under various applied voltages; the blue zone represents the conditions where droplets could stably generated while pink zone represents the conditions where no/unstable droplets could be generated. If V_{oil} is smaller than $4 V_{pp}$ (or the V_{water} is smaller than $2.5 V_{pp}$), the induced driving force for pumping oil (or for pumping water) is too small to overcome the friction and droplet cannot be generated. If the V_{water} is as large as $6 V_{pp}$, droplets cannot be generated regardless of the V_{oil} as the oil stream cannot cut/squeeze a vigorous water stream, leading to the formation of concurrent flow. In the case of $V_{water} = 2.5 - 4.0 V_{pp}$, the droplet size decreased from the sizes larger than the channel ($100 \mu m$) to the smaller sizes when V_{oil} was increased, indicating that the mechanism of droplet generation was transitioning from the squeeze regime to the shear regime as the V_{oil} increases (Garstecki et al. 2006). The LCAT droplet generator was able to generate W/O droplets as small as $50 \mu m$ with generation frequency ~ 21 Hz when the V_{oil} and V_{water} were set to $16 V_{pp}$ and $3.5 V_{pp}$, respectively. The maximum generation frequency was 31 Hz (the droplet diameter is around $175 \mu m$) when the V_{oil} and V_{water} were set to 16 and $5.0 V_{pp}$, respectively.

We summarize the correlation of W/O droplet size to the applied voltages (V_{oil} and V_{water}) as shown in figure 9. If the V_{water} was fixed at $4.0 V_{pp}$, the maximum controllable range of droplet size ($55-360 \mu m$) could be obtained when V_{oil} was between 6 to $16 V_{pp}$. On the other hand, if the V_{oil} was fixed at $12 V_{pp}$, the maximum controllable range of droplet size (60 to $420 \mu m$) could be obtained when V_{water} was between 3.5 to $5.5 V_{pp}$. A variety of combinations of V_{oil} and V_{water} can be utilized to generate a certain droplet size. For example, any combinations of V_{oil} / V_{water} such as $5.2 / 3.0$, $7.0 / 3.5$, $9.2 / 4.0$, $10.8 / 4.5$, or $13.8 / 5.0 (V_{pp} / V_{pp})$ will generate $200 \mu m$ droplets (see a blue dash line). For these combinations, the generated droplets were similar in size, but the corresponding generation frequencies are very different, namely 2.1, 5.5, 8.9, 9.6, and 16.0 Hz (for $5.2 / 3.0$, $7.0 / 3.5$, $9.2 / 4.0$, $10.8 / 4.5$, and $13.8 / 5.0 V_{pp} / V_{pp}$), respectively. Evidently, increasing V_{oil} and V_{water} increased the generation frequency of droplets.

O/W droplet generation

Figure 10 shows the O/W droplet generation with the LCAT droplet generator. Under a fixed V_{oil} , the size and generation frequency of O/W droplets increased with V_{water} . With the LCAT droplet generator, the generated oil droplets could be as small as $60 \mu m$ with generation frequency approximately 26.3 Hz when the V_{oil} and V_{water} were 4 and $6 V_{pp}$, respectively. The maximum generation frequency could be up to 59.1 Hz (the droplet diameter is around $83 \mu m$) when the V_{oil} and V_{water} were 7 and $6 V_{pp}$, respectively. The controllable range of O/W droplet size was $60-150 \mu m$ (see figure 11), narrower than that of W/O droplet generation ($55-420 \mu m$). Regarding O/W droplet generation system, the flow rate of continuous phase (water) was larger than that of dispersed phase (oil), resulting in a relatively large shear force that is the primary mode of droplet generation. Due to the droplet

generation being in the shear regime, the O/W droplets were generally smaller or slightly larger than 100 μm . In some cases (e.g. $V_{oil}/V_{water} = 5/2$ or $6/2$), the droplet size was around 150 μm larger than the channel width, which meant the droplet was squeeze-generated. On the other hand, for the W/O droplet generation system, the flow rate of continuous phase (oil) was usually smaller than that of dispersed phase (water). The droplet generation was therefore mainly under the squeeze regime where a wider controllable range of droplet size can be achieved.

Conclusion

In this work, we successfully demonstrate an LCAT droplet generator with integrated LCAT pumps for the generation of microfluidic O/W or W/O droplets. The geometric optimization of LCAT pump revealed that the performance improved with an increase in the width of junction between side and main channel (a) and a decrease in the length of side channel (c). The larger junction width the pump has, the more acoustic energy the pump can effectively transfer to produce a stronger microstreaming for pumping liquid. By means of FDTD vapor deposition on the channel surface, stable and long lasting air pockets trapped inside side channels can be formed to allow acoustic-induced functional microstreaming to pump oil liquid. We summarized the droplet generation with our LCAT droplet generator in terms of size and generation frequency. For the W/O droplet generation, the controllable range of droplet diameter is 50–420 μm and the maximum generation frequency was 31 Hz (175 μm diameter droplet), while for the O/W droplet generation, the controllable range of droplet diameter is 60–150 μm and the maximum generation frequency was 59 Hz (83 μm diameter droplet). The LCAT droplet generator possesses integrated and valveless pumps that can be accurately controlled by the acoustic source (e.g. piezoelectric transducer, PZT). The voltage applied to generate microfluidic droplets can be as low as 4 V_{pp} . Because of the simple design without the need for external pumps, its fabrication is facile and low cost. Furthermore, the LCAT droplet generator can be easily integrated with a wide range of LCAT on-chip microfluidic devices (e.g. pump, mixer, switch, sorter, etc.) to enable portable and fieldable devices for point-of-care diagnostics in remote and low resource settings.

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Fig. 1.
Schematics of LCAT droplet generator and configuration of operation.

Fig. 2.
Five different LCAT devices (D1, D2, D3, D4, and D5).

Fig. 3.
Schematic depiction for measurement of performance of the LCAT pumps.

Fig. 4. Dependence of flow rate (Q) on differential pressure (P) for five different LCAT pumps. The number of independent measurements was 5 ($N=5$).

Fig. 5. Liquid-air oscillation of various LCAT pumps nearby the junction of main and side channels (applied voltage = 3.0 V_{pp}, 67.7 kHz).

Fig. 6. Contact angles of water and oil drops on PDMS surfaces with (lower-left) and without (upper-left) FDOTS surface treatment, and formation of liquid-gas interfaces in channels of LCAT pumps with (lower-right) and without FDOTS surface treatment (upper-right). The number of independent measurements was 5 ($N=5$).

Fig. 7.

The correlations between maximum back pressure (P_{max}) and applied voltage for the D4 pumping water and mineral respectively. The number of independent measurements was 5 (N=5).

Fig. 8.

W/O droplet generation under various applied voltages (V_{oil} / V_{water}). The scale bar indicates 200 μm . The number of independent measurements was 5 ($N=5$).

Fig. 9.

The correlation among W/O droplet diameter and applied voltages (V_{oil} and V_{water}). The number of independent measurements were 5 ($N=5$).

V_{oil} / V_{water}
Generation frequency (Hz)
Droplet diameter (μm)

Fig. 10.

O/W droplet generation under various applied voltages (V_{oil} / V_{water}). The scale bar indicates 200 μm . The number of independent measurements was 5 ($N=5$).

Fig. 11.

The correlation among O/W droplet diameter and applied voltages (V_{oil} and V_{water}). The number of independent measurements was 5 ($N=5$).