

# Three-Phase ScAlN-based PMUT driven Acoustic Steaming Micropump

Chen WU

wu86@imec.be

KU Leuven <https://orcid.org/0000-0003-3547-1577>

Grim Keulemans

Imec

Benjamin Jones

Imec <https://orcid.org/0009-0004-6139-0484>

Xavier Rottenberg

Imec

Veronique Rochus

Imec

Paul Heremans

Interuniversity Microelectronics Centre

---

## Article

### Keywords:

**Posted Date:** September 15th, 2025

**DOI:** <https://doi.org/10.21203/rs.3.rs-7170127/v1>

**License:**   This work is licensed under a Creative Commons Attribution 4.0 International License.  
[Read Full License](#)

**Additional Declarations:** There is no conflict of interest

---

**Version of Record:** A version of this preprint was published at Microsystems & Nanoengineering on May 26th, 2026. See the published version at <https://doi.org/10.1038/s41378-026-01339-5>.

# Abstract

Scandium-doped aluminum nitride (ScAlN)-based piezoelectric micromachined ultrasonic transducer (PMUT) arrays have drawn increasing attention in acoustofluidics for micro total analysis systems ( $\mu$ TAS), primarily in applications involving acoustic radiation force for bioparticles and cell manipulations. However, their use for fluid handling via acoustic streaming remains underexplored. This study, for the first time, examines the potential of a rectangular membrane ScAlN-based PMUT array to generate directional acoustic streaming for micro pumping applications. The PMUT array is embedded within a PDMS microfluidic channel and is applied by a set of AC signals with a phase difference of  $120^\circ$  to adjacent PMUT cells to induce directional streaming flow. The device features a compact active area of 1.2 mm x 1.6 mm and demonstrates a volumetric flow rate of 0.1  $\mu$ L/min, in good agreements with predictions from numerical multiphysics simulations. Further numerical optimization suggests that the flow rates of 1.0  $\mu$ L/min are achievable by optimizing the array kerf and applied phase difference to adjacent PMUTs. A comparative analysis with state-of-the-art chip integrable micropumps highlights the advantages of the proposed device, including its miniaturized footprint, CMOS compatibility, and ease of on-chip integration. These attributes position the proposed micropump as a promising solution for  $\mu$ TAS applications, especially where compact size and precise, low-flow-rate fluid control are critical.

## Introduction

On-chip microfluidics is a core enabling technology for lab-on-a-chip (LOC) or micro total analysis systems ( $\mu$ TAS), supporting essential functions such as fluid mixing, pumping, particle manipulation, cell separation, and DNA transportation. Various on-chip microfluidic actuators have been developed, relying on different physical mechanisms. Mechanical actuators, including pneumatic [1] and piezoelectric micropumps [2], manipulate fluid via pressure differentials induced by membrane deformation. Dielectrophoretic actuators utilize polarization forces in non-uniform electric fields and are widely applied for blood plasma separation [3] and continuous cell sorting [4], [5]. Electroosmotic pumps, driven by ion motion in the electrical double layer near charged channel walls, offer fine control over flow rates and reagent injection for cell culture applications [6]. Thermoelectric devices create temperature gradients under applied voltages, which can halt fluid flow or enable on-chip single cell cryopreservation [7]. Optical tweezers [8] are used for single cell trapping and spectroscopy. Magnetophoretic devices, which operate under two- or three-dimensional (2D or 3D) magnetic fields, enable the precise control of magnetic particles and magnetically labelled cells [9], [10].

Apart from the above-mentioned devices, acoustofluidic devices, which utilize acoustic waves for manipulation of fluids, have gained significant attention for their contactless operations and biocompatibility. They are widely applied in bioparticle manipulation, single-molecule analysis, tissue engineering, and point-of-care diagnostics [11]. Acoustic waves offer a broad operational frequency range (kHz to GHz), allowing for the manipulation of particles of varying sizes through two key forces: acoustic radiation force (scales with particle volume) and acoustic streaming drag force (scales with particle diameter). A critical particle size distinguishes the dominant force of influence and is dependent

on the acoustic wavelength and intensity [12]. For example, the critical particle diameter is determined to be around 2  $\mu\text{m}$  for a spherical polystyrene particle suspended in water at an acoustic frequency around 2 MHz [13]. Consequently, large biological particles and cells are typically manipulated by radiation forces, while sub-micron particles are primarily affected by the induced drag forces via the acoustic streaming motion of the fluid. Acoustofluidic devices are thus designed to harness one or both forces for targeted manipulation.

Acoustic transducers form the foundation of acoustofluidic devices and can be classified into surface acoustic wave (SAW), bulk acoustic wave (BAW), and micromachined ultrasonic transducers (MUTs). SAW devices typically use interdigitated transducers (IDTs) composed of comb-like electrode fingers on a piezoelectric substrate. These devices are widely used for droplet transport [14], pumping [15], 1D or 2D cell patterning [16], and particle manipulation [17], [18]. BAW transducers such as solidly mounted resonators (SMRs), operate by propagating acoustic energy through the substrate and fluid. SMR's can be used to generate high intensity acoustic beams; for example, You *et al.* demonstrated a GHz frequency Eckart acoustic streaming micropump for drug delivery purposes [19].

Compared to conventional SAW and BAW transducers, MUTs including capacitive MUTs (CMUTs) and piezoelectric MUTs (PMUTs) offer several advantages. They offer size miniaturization and the potential for on-chip integration with other components for  $\mu\text{TAS}$  systems. Their microscale free-standing membranes enable large displacement amplitudes at lower operating frequencies in a small form factor making them well-suited for compact acoustofluidic devices. In a CMUT, the membrane includes a metal conductive layer, and the conductive substrate serves as the bottom electrode [20]. A DC bias voltage pre-stresses the membrane due to the electrostatic force while an additional AC voltage generates acoustic waves that propagate through the surrounding medium. A PMUT consists of a thin-film piezoelectric layer sandwiched between top and bottom electrodes [21]. The membrane vibrates when an AC signal is applied between the electrodes, eliminating the high DC bias needed for CMUT devices which could be problematic for living cells in microfluidics, making PMUT a more biosafe solution.

PMUTs based on aluminum nitride (ALN) have shown strong potential for on-chip applications. ALN is a CMOS compatible, biocompatible, lead-free material, and exhibits high thermal conductivity, low dielectric and acoustic loss, and good stability, making it ideal for high-frequency actuators. Recent works utilizing ALN PMUT arrays include Qian *et al.* (for multisite particle manipulation) [22] and Li *et al.* (for bubble-based stirring and particle patterning) [23]. Furthermore, by combining both acoustophoretic and dielectrophoretic forces, Weekers *et al.* demonstrated dynamic T-cell manipulations using 9.5% Sc-doped ALN PMUT actuators [24].

Past studies using PMUT arrays primarily use acoustic radiation forces for manipulations; however, the potential of acoustic streaming generated by these arrays remains under-investigated. In this work, we experimentally demonstrate streaming induced by a three-phase ScAlN-based PMUT array for micropump applications. The compact micropump features embedded actuators within a microfluidic channel. Numerical simulations, which are validated against the experimental results, are used to predict

how the design can be optimized for flow rates suitable for certain  $\mu$ TAS applications. Benchmarking to state-of-the-art micropumps from the literature indicate the proposed design compares favorably in terms of miniaturization, CMOS compatibility, and ease of on-chip integration.

## Results

### Pumping mechanism

The cross-sectional schematic of the fabricated micropump device is shown in Fig. 1. A thin film ScAlN piezoelectric layer is sandwiched between a top and bottom electrode to form a free-standing membrane. By applying an AC signal to the top electrode while grounding the bottom electrode, the electrical excitation can be converted into an oscillatory mechanical vibration, generating an acoustic wave that propagates away from the PMUT. Note that the acoustic wave generated by a single PMUT produces no net flow. Therefore, three sets of separate AC signals of the same frequency but offset phase ( $\Delta\varphi = 120^\circ$ ) are applied to adjacent PMUTs within the array (as shown in Fig. 1) to generate a propagating acoustic wave along the length of the channel.

### Prediction of streaming velocity of the micropump

The streaming velocity of the micropump was predicted via Finite Element Method (FEM) numerical simulations. As illustrated in Fig. 2, the model consists of one PMUT unit cell with cyclic periodic boundary conditions at a  $120^\circ$  phase difference applied between the left and righthand sides of the domain and the laminar fluid flow equations are solved with periodic conditions applied to the inlet (lefthand side) and outlet (righthand side) of the microfluidic channel ( $\Gamma$  in Fig. 2). The width of the free-standing membrane is  $50\text{ }\mu\text{m}$ , and the kerf, which is the distance between adjacent PMUT membranes is  $50\text{ }\mu\text{m}$ . The PDMS channel top wall (2 mm thick) is treated as a linear elastic material with a shear viscosity of  $3.5\text{ Pa}\cdot\text{s}$ , a longitudinal wave velocity of  $1030\text{ m/s}$  and a shear wave velocity of  $100\text{ m/s}$  [25]. The inner top electrode has a length of  $32\text{ }\mu\text{m}$ . The floating outer electrodes ( $6\text{ }\mu\text{m}$ , at a distance of  $3\text{ }\mu\text{m}$  of the inner electrodes) can help suppress spurious vibrations and enhance displacement amplitude. Further details of the numerical model are given in the Materials and Methods section. Different components as well as their materials and thickness are listed in Table 1.

Table 1  
Dimensions and parameters used in FEM model

| Component            | Material                       | Thickness |
|----------------------|--------------------------------|-----------|
| Substrate            | SiO <sub>2</sub>               | 4 μm      |
| Piezoelectric layer  | ScAlN (9.5% Sc doping)         | 1.3 μm    |
| Passivation          | Si <sub>3</sub> N <sub>4</sub> | 1.5 μm    |
| Electrode            | Al                             | 0.4 μm    |
| Cavity               | Vacuum                         | 0.6 μm    |
| Microfluidic channel | Water                          | 360 μm    |
| Channel top wall     | PDMS                           | 2 mm      |

A frequency sweep was conducted to extract the resonant frequency of one third of the periodic pumping section, which is 4.37 MHz under water. The maximum out-of-plane displacement of the PMUT membrane is 0.55 nm/V<sub>PP</sub> under resonance. The PMUT is then driven at 4.37 MHz to numerically evaluate the streaming pattern inside the microfluidic channel for two cases: (1) all PMUT's driven in-phase ( $\Delta \varphi = 0^\circ$ ) and (2) adjacent PMUT driven with a phase difference of  $\Delta \varphi = 120^\circ$  (see Fig. 3 (a) and (b), respectively).

The streamlines are symmetric about the PMUT centreline ( $X = 0$ ) when  $\Delta \varphi = 0^\circ$ , as observed in Fig. 3 (a); therefore, no net flow. In comparison, by applying a phase difference of  $120^\circ$ , the symmetry is broken, leading to a directional streaming flow shown in Fig. 3 (b). In both cases, vortices are observed in proximity to the corners of the PMUT membrane, due to the discontinuity in vibrations along the actuator surface as only the PMUT membrane generates substantial displacements.

### Device description

As shown in Figs. 1 and 4, the PMUT array micropump consists of  $9 \times 1$  rectangular PMUT unit cells. The thicknesses of different layers are listed in Table 1. The experimental PMUT array had a cell pitch of 100 μm and a vibrational membrane width of 50 μm. The chip was wire bonded to a printed circuit board (PCB) for electrical routing. A phase difference of  $120^\circ$  was applied to adjacent PMUT cells inside the array. As a result, each third PMUT was driven by the same signal. A polydimethylsiloxane (PDMS) microfluidic channel (length = 3.6 mm, width = 1.6 mm, and height = 0.36 mm) was plasma bonded on top of the PMUT membrane. The micropump device (including the PMUT actuator die and PDMS microfluidic channel) was completely submerged in liquid within a PMMA reservoir to allow the fluid to recirculate after exiting the microfluidic channel (depicted in Fig. 4).

### Device Characterization

The resonant frequency of the device inside the microfluidic channel filled with water is 4.3 MHz was measured by Laser Doppler Vibrometer (LDV), which is very close to simulation results (4.37 MHz). However, when driving the PMUT array at the resonant frequency, due to the large surface area of the rectangular PMUT, the acoustic crosstalk between adjacent PMUT cells is significant. As a result, obtaining the desired displacement phase difference between adjacent PMUT cells ( $\Delta \varphi = 120^\circ$ ) is challenging. Therefore, the micropump is driven off-resonance at 3.92 MHz to reduce the acoustic coupling and obtain a displacement phase difference closer to intended. The maximum out-of-plane displacement magnitude is 0.46 nm/V<sub>pp</sub> (averaged over 9 PMUTs). The measured displacement phase difference between adjacent cells is 120° +/- 36°. The simulated instantaneous PMUT membrane displacement (driven at 4.37 MHz) was compared with LDV measurements (driven under 3.92 MHz), and a good match was identified with measured displacement amplitude approximately 10% lower than simulations. For a more detailed comparison, see Fig. 2 in Supplementary section 1.

### Characterization of Streaming Velocity

Particle tracking measurements were conducted to assess fluid flow during actuation of the PMUT-based micropump. The device was driven under 3.92 MHz and 40 V<sub>pp</sub>, these conditions were used for all following discussions. Fluorescent particles were imaged in the PMUT active section (see Fig. 4) at two z positions, 180 µm (mid-depth of channel) and 20 µm away from surface of the actuator device. Overlaid images constructed from particle images taken at two separate points are shown in Fig. 5. Particles labeled green represent the initial particle position while the red-labelled particles represent the position after 30 seconds have elapsed. Forward flow is observed at mid-depth (Fig. 5 (a)) while reverse flow is observed close to channel bottom (Fig. 5 (b)). Particles labeled yellow appear to be immobilized on the surface of the actuator device (produced by an overlay of the green and red particles images).

The simulated streaming velocity was evaluated at the outlet of the microfluidic channel (right hand side (RHS) boundary of the simulation model shown in Fig. 2) along the channel depth, solid line shown in Fig. 6. The maximum flow velocity achieved is 0.51 mm/min, with a mean streaming velocity of 0.35 mm/min.

The measured fluid velocity as a function of the position along the channel height was compared to simulation results, dashed lines shown in Fig. 6. Measurements were taken at both the inlet and PMUT array active section (see Fig. 4 for measurement locations) and indicated a maximum flow velocity of 0.34 and 0.42 mm/min at the inlet and active sections, respectively, and a mean flow velocity of 0.16 and 0.15 mm/min at the respective measurement locations. The simulated fluid velocity is in good agreement with measurements. Vortices are also noted in the experimental measurements, as indicated by the reverse flow measured close to the actuator surface in both Figs. 5 and 6. Given the micropump's cross section is 1.6 mm × 0.36 mm, the resulting volumetric flow rate is approximately 0.1 µL/min, as estimated from measured fluid velocity profile, with an applied voltage amplitude of 20 V compared to 0.2 µL/min as simulated.

## Optimization of the Flow Rate via Parameter Sweep Analysis

We investigated the impact of two PMUT array parameters on the streaming velocity, which are the array kerf (depicted in Fig. 1) and the phase difference in applied AC signals between adjacent PMUT unit cells ( $\Delta\varphi$ ). We applied the same FEM model presented above (Fig. 2) that demonstrated a good match in acoustic streaming flow pattern, fluid velocity, and the volumetric flow rate between numerical study and particle tracking experiments. The microfluidic channel height was kept fixed at 360  $\mu\text{m}$  (same as used in previous simulations and experiments). With cyclic periodic boundary conditions applied to both sides, the phase difference  $\Delta\varphi$  was swept from 50° to 180° with a step of 10° for 5 different fixed values of the kerf, ranging from 10  $\mu\text{m}$  to 50  $\mu\text{m}$  with a step of 10  $\mu\text{m}$ . The resonant frequency of the periodic device and the volumetric flow rates under resonance were evaluated as shown in Fig. 7 (a) and (b).

The resonance of the periodic device increases as  $\Delta\varphi$  increases and as kerf decreases. Notably, even with the same PMUT membrane dimensions and material characteristics, the resonance frequency of the periodic device changes significantly as the kerf and/or phase difference is altered. This is because the periodic device resonance is not merely determined by the PMUT itself, but a resultant of multiple couplings: acoustic crosstalk between PMUTs embedded inside the microfluidic channel filled with water and interference due to the reflected acoustic wave from the PDMS top wall. The maximum achievable flow rate increases with smaller kerf, while this optimal flow rate occurs at lower phase difference. The maximum flow rate is identified for an array kerf of 10  $\mu\text{m}$  and a phase difference of 70°, leading to an enhancement in flow rate by a factor of 6 compared to the simulation result of the tested design (kerf of 50  $\mu\text{m}$  and  $\Delta\varphi = 120^\circ$ ). However, a phase difference of 70° requires the periodic pumping section to be composed of 36 PMUT cells, all driven at a different phase, making the design and implementation of the driving circuitry complicated. Reducing the number of PMUT cells to 4 within a periodic pumping section (i.e., phase difference of 90°) with an array kerf of 20  $\mu\text{m}$  gives a slightly reduced enhancement in flow rate by a factor of 5 compared to the tested design, representing, perhaps, a more practical and easier to implement design. In the end, we project an increase in flow rate from 0.2  $\mu\text{L}/\text{min}$  to 1.0  $\mu\text{L}/\text{min}$  when using an optimized four-phase design versus the simulated three-phase device.

## Discussion

Table 2 presents a comparison between the proposed micropump and various state-of-the-art chip-integrable micropumps described elsewhere in the literature. While the proposed device exhibits a relatively modest flow rate under a moderate input voltage (40 Vpp), it stands out for its compact footprint and CMOS compatibility.

Table 2

Performance comparison between the state-of-the-art on-chip micropumps and the proposed device

| Reference                        | Micropump type                                                    | Channel dimension $L \times W \times H$ (mm <sup>3</sup> ) | Input voltage or power    | Flow rate (μL/min) | CMOS compatibility       |
|----------------------------------|-------------------------------------------------------------------|------------------------------------------------------------|---------------------------|--------------------|--------------------------|
| Holman <i>et al.</i> 2023 [26]   | Piezoelectric diaphragm micropump + check valves                  | 12 mm in diameter, 0.2 mm in height                        | 30–140 V <sub>pp</sub> AC | 3.4–41.8           | No (PZT)                 |
| Afrasiab <i>et al.</i> 2011 [27] | Valveless travelling flexural plate wave micropump                | $5.2 \times 0.2 \times 0.3$                                | 40 V <sub>pp</sub> AC     | 407.4              | No (PZT)                 |
| Okamoto <i>et al.</i> 2020 [28]  | Electro-osmotic micropump                                         | $7.5 \times 0.4 \times 0.05$                               | 5 V DC                    | 0.167              | Yes                      |
| Chen <i>et al.</i> 2019 [29]     | Micropores evaporation driven micropump                           | Large but not specified, $>10 \times 10 \times 0.25$       |                           | 0.24               | No                       |
| Wang <i>et al.</i> 2017 [15]     | IDT-based SAW micropump                                           | $10 \times 0.5 \times 0.07$                                | 7.5 W                     | 0.08               | No (LiNbO <sub>3</sub> ) |
| You <i>et al.</i> 2024 [30]      | GHz Eckart acoustic streaming micropump                           | $9 \times 9 \times 9$                                      | 1 W                       | 11800              | No (SMR)                 |
| Zhang <i>et al.</i> 2025 [31]    | Sharp edge micropillar-based acoustic streaming micropump induced | $25 \times 30 \times 0.5$                                  | 54 V <sub>pp</sub> AC     | 16.2               | No (PZT)                 |
| Gao <i>et al.</i> 2020 [32]      | Air bubble-based acoustic streaming micropump                     | $75 \times 0.8 \times 0.1$                                 | 7 V <sub>pp</sub> AC      | 1.6                | No (PZT)                 |
| <b>This work</b>                 | PMUT-based acoustic streaming micropump                           | $3.6 \times 1.6 \times 0.36$                               | 40 V <sub>pp</sub> AC     | 0.1–1.0            | Yes (ScAlN)              |

Traditional piezoelectric diaphragm micropumps (as an example, see Holman *et al.* [26]) are typically large and rely on PZT, which is not a material compatible with typical CMOS foundries. Moreover, the typical inclusion of mechanical check valves increases complexity and introduces additional hydraulic resistance. Afrasiab *et al.* [26] demonstrated a high flow rate generated by a valveless flexural plate wave (FPW) micropumps. However, they also use PZT as the piezoelectric material for the membranes, limiting CMOS integration possibilities. Furthermore, the performance of FPW micropumps as described by [27] is highly dependent on the coupling between membrane and channel dimensions, which

constrains design flexibility. The electro-osmotic micropump developed by Okamoto *et al.* [28] is CMOS compatible; however, it requires high voltages (up to 100 V). Chen *et al.* [29] developed a passive micropump driven by the evaporation of liquid within micropores exposed to the surrounding atmosphere; thus no external power source is required but the sensitivity to environmental conditions and lack of on-demand flow control limits the utility. SAW-driven micropumps, such as the one developed by Wang *et al.* [15], produce low displacement amplitudes, which typically produce low flow rates. As an additional drawback, the SAW IDT actuators are usually external to the microfluidic channel resulting in severe acoustic attenuation through the channel side wall, reducing the effectiveness of the device to pump fluid. You *et al.* [30] developed a micropump based on GHz-frequency Eckart streaming, which produced high flow rates through localized, intense acoustic beams. However, the SMR actuator used is not readily embedded within a monolithic microfluidic system. These systems also require additional components, such as, impedance matching circuits, and nozzle-diffuser geometries to guide the flow, which results in increased design complexity. Similarly, micropumps that use sharp edge micropillars (Zhang *et al.* [31]) or acoustic air bubbles (Gao *et al.* [32]) rely on external PZT transducers and bulky structures, making them incompatible for fully integrated CMOS-based devices. In contrast, the proposed ScAlN-based PMUT micropump offers significant advantages: it is miniaturized, CMOS compatible, and easily integrated on-chip. Its performance aligns well with the requirements of a  $\mu$ TAS, especially where compact size and precise, low-flow-rate fluid control are critical.

Although our proposed device demonstrated a limit in the maximum supported flow rate, the pumping performance is appropriate for certain  $\mu$ TAS applications reported in the literature, where the flow rates can fall within a wide range depending on the testing requirements. For instance, sub-microliter per minute flow rates are sufficient for sweat analysis systems [29] while between 1 and 10  $\mu$ L/min are needed in some cell culture chips [26]. Applications requiring higher flow rates include blood cell separation platforms [33], which can be in the hundred  $\mu$ L/min range, and wildlife toxicity testing [34] requiring up to a thousand  $\mu$ L/min.

To extend the capability of the micropump to other application domains, improvements in the PMUT membrane displacement are necessary. This can be achieved by increasing the Sc doping concentration in ScAlN to around 30%, which enhances the piezoelectric coefficient. Also, reducing the thickness of the suspended PMUT membrane can further increase its mechanical displacement and overall efficiency. Further work should also include full 3D simulations to capture more realistic boundary effects, particularly the influence of clamped edge conditions and membrane geometry. While vortex formation may hinder directional streaming in micropump applications, it could be leveraged for efficient micro mixing. This dual functionality presents an opportunity for expanded use of this device in  $\mu$ TAS applications.

## Conclusion

This work explores the potential of acoustic streaming generated by a three-phase PMUT array for micropump applications. We demonstrate that applying a phase difference to adjacent PMUT elements

induces net fluid flow, enabling controlled fluid transport in acoustofluidic systems. Our results show a steady laminar flow in the bulk fluid region, although localized recirculation and vortex formation are observed near the membrane. Numerical predictions closely match particle tracking experimental results, validating the simulation approach. Further numerical analysis reveals that the pumping performance can be significantly enhanced by optimizing the PMUT array's kerf spacing and phase shift. For example, an optimized four-phase configuration projects a fivefold increase in flow rate. The proposed micropump is compact, CMOS-compatible and fully integrable on-chip, making it well-suited for applications where precise control of low-volume fluid flow is critical.

To conclude, our study demonstrates a new capability of PMUT arrays to generate acoustic streaming within a fluid and produce a net flow suitable for microfluidic pumping. Our combined theoretical and experimental insights provide a foundation for future development of PMUT-driven acoustic streaming devices for microfluidic systems.

## **Materials and Methods**

### **Device microfabrication**

A silicon-based PMUT array with 9.5% Sc-doped AlN piezoelectric material was fabricated using standard semiconductor manufacturing techniques using 200 mm wafers at SilTerra Malaysia Sdn. Bhd. The microfluidic channel was fabricated in PDMS using soft lithography technique. The master mold was 3D printed by Proto Labs Germany GmbH using an ABS-like MicroFine™ material (Proto Labs, Inc., USA). Inlet and outlet reservoirs were designed at the terminals of the microfluidic channel with a diameter of 4 mm. The PDMS, a mixture of elastomer base (SYLGARD 184, Dow Corning, USA) at a 10:1 ratio by mass between resin and curing agent, was centrifugated at 0.3 rcf for 3 minutes at room temperature, poured over the master, and cured for 2 hours at 100 °C and 950 mbar. The PDMS was peeled off, and subsequently, plasma bonded on top of the PMUT array. The device was incubated at 100 °C for 3 hours to achieve a strong bonding between the PDMS and PMUT array membrane. The PMMA reservoir was fabricated via laser cutting.

### **Experimental Setup**

The microfluidic channel as well as the PMMA reservoir were prefilled with highly purified water (HPW). A frequency sweep was conducted with a Laser Doppler Vibrometer (LDV) (MSA500, Polytec, Germany) to extract the resonant frequency of each individual PMUT cell. The PMUT displacement amplitude as well as the phase were characterized and compared with simulation results. The microfluidic channel as well as the PMMA reservoir were loaded with a 1.2 µm diameter fluorescent particle suspension. The streaming velocity profile is assessed by particle tracking under a fluorescent microscope (Zeiss Axio Examiner Z1, Germany) within two separate measurement sections, inlet and PMUT active sections respectively, depicted in Fig. 4.

### **Finite Element Method (FEM) Numerical Simulation**

A fully coupled 2D FEM model of one third of a periodic pumping section, containing one PMUT unit cell, has been created in COMSOL version 6.2. The simulation incorporates electrostatics and solid mechanics for the PMUT array, and the thermal viscous acoustics for the microfluidic channel, which is filled with water. The “piezoelectric effect” and the “acoustic-structure boundary” multiphysics interfaces are employed to solve the first-order acoustics equations (3) and (4) in supplementary section 2, yielding the primary acoustic velocity  $\vec{v}_1$  and density  $\rho_1$ . Intermediate calculations are performed to determine the streaming contributions, including the mass source and body force. Subsequently, the “acoustic streaming domain coupling” and “acoustic streaming boundary coupling” multiphysics interfaces are used within the laminar flow module to solve for the second-order acoustic streaming equations (5) and (6) in supplementary section 2, generating the streaming velocities in the fluid domain. A mesh convergence analysis was performed, and a mesh density corresponding to approximately 170 elements per acoustic wavelength was determined to be sufficient for accurate simulation results (further details on the implemented mesh can be found in Supplementary section 3).

## Declarations

### Acknowledgment

The authors thank Pieter Gijsenbergh for designing PCB's, Chad Arnett for assistance in PDMS microfluidic channel fabrication, Guilherme Brondani Torri for advice and insightful discussions and other members within the fluidic and MEMS group at Imec for their kind help and contributions.

### Conflict of Interests

The authors declare no competing interests.

### Author Contributions

CW: silicon microfabrication of test devices, design and fabrication of experimental setup, conducting experiments and characterizations, formulation and development of numerical model, analysis of results, and drafting and editing the manuscript.

GK: guidance and support for the development of experimental setup, numerical model, results and analysis and support for the drafting of the manuscript and reviewing/editing.

BJ: guidance and support for the development of experimental setup, numerical model, results and analysis and support for the drafting of the manuscript and reviewing/editing.

VR: manuscript reviewing and editing

XR: initial device concept, advice supporting various technical and scientific aspects of the work, and manuscript reviewing and editing

PH: advice supporting various technical and scientific aspects of the work and manuscript reviewing and editing

**Supplementary information** accompanies the manuscript on the *Microsystems & Nanoengineering* website <http://www.nature.com/micronano>.

## References

1. Clime, L., Brassard, D., Geissler, M. & Veres, T. Active pneumatic control of centrifugal microfluidic flows for lab-on-a-chip applications. *Lab Chip* 15, 2400–2411 (2015).
2. Zhao, B., Li, X., Shi, J. & Liu, H. Design and Experimental Characterization of a Microfluidic Piezoelectric Pump Utilizing P(VDF-TrFE) Film. *Coatings* 14, 1483 (2024).
3. Srivastava, S. K., Artemiou, A. & Minerick, A. R. Direct current insulator-based dielectrophoretic characterization of erythrocytes: ABO-Rh human blood typing. *ELECTROPHORESIS* 32, 2530–2540 (2011).
4. Alazzam, A., Mathew, B. & Alhammadi, F. Novel microfluidic device for the continuous separation of cancer cells using dielectrophoresis. *Journal of Separation Science* 40, 1193–1200 (2017).
5. Li, H. & Bashir, R. Dielectrophoretic separation and manipulation of live and heat-treated cells of *Listeria* on microfabricated devices with interdigitated electrodes. *Sensors and Actuators B: Chemical* 86, 215–221 (2002).
6. Glawdel, T. & Ren, C. L. Electro-osmotic flow control for living cell analysis in microfluidic PDMS chips. *Mechanics Research Communications* 36, 75–81 (2009).
7. Sgro, A. E., Allen, P. B. & Chiu, D. T. Thermoelectric Manipulation of Aqueous Droplets in Microfluidic Devices. *Anal. Chem.* 79, 4845–4851 (2007).
8. Liberale, C. et al. Integrated microfluidic device for single-cell trapping and spectroscopy. *Sci Rep* 3, 1258 (2013).
9. Abedini-Nassab, R. & Shourabi, R. High-throughput precise particle transport at single-particle resolution in a three-dimensional magnetic field for highly sensitive bio-detection. *Sci Rep* 12, 6380 (2022).
10. Abedini-Nassab, R., Murdoch, D. M., Kim, C. & Yellen, B. B. Optimization of magnetic switches for single particle and cell transport. *Journal of Applied Physics* 115, 244509 (2014).
11. Rufo, J., Cai, F., Friend, J., Wiklund, M. & Huang, T. Acoustofluidics for biomedical applications. *Nature Reviews Methods Primers* 2, 30 (2022).
12. Barnkob, R., Augustsson, P., Laurell, T. & Bruus, H. Acoustic radiation- and streaming-induced microparticle velocities determined by micro-PIV in an ultrasound symmetry plane. *Phys. Rev. E* 86, 056307 (2012).
13. Muller, P., Barnkob, R., Jensen, M. & Bruus, H. A numerical study of microparticle acoustophoresis driven by acoustic radiation forces and streaming-induced drag forces. *Lab on a chip* 12, 4617–27 (2012).

14. Jue, B. et al. Manipulating acoustic streaming in a droplet using a single surface acoustic wave transducer. *Applied Acoustics* 231, 110517 (2025).
15. Wang, T., Ni, Q., Crane, N. & Guldiken, R. Surface acoustic wave based pumping in a microchannel. *Microsystem Technologies* 23, 1–8 (2017).
16. Ding, X. et al. Tunable patterning of microparticles and cells using standing surface acoustic waves. *Lab on a chip* 12, 2491–7 (2012).
17. Stringer Martin, M. Acoustofluidic manipulation of microparticles using surface acoustic wave induced streaming. (Cardiff University, 2024).
18. Ding, X. et al. On-chip manipulation of single microparticles, cells, and organisms using surface acoustic waves. *Proceedings of the National Academy of Sciences of the United States of America* 109, 11105–9 (2012).
19. You, R. et al. A Miniaturized Wireless Micropump Enabled by Confined Acoustic Streaming. *Research* 7, (2024).
20. Khuri-Yakub, P. & Oralkan, O. Capacitive micromachined ultrasonic transducers for medical imaging and therapy. *Journal of micromechanics and microengineering: structures, devices, and systems* 21, 54004–54014 (2011).
21. Moisello, E. et al. PMUT and CMUT Devices for Biomedical Applications: A Review. *IEEE Access* 12, 18640–18657 (2024).
22. Qian, J. et al. Dual-Function Multichannel Acoustofluidic Particle Manipulation Enabled by a PMUT Array. *IEEE Transactions on Electron Devices* PP, 1–7 (2024).
23. Li, X. et al. Piezoelectric Micromachined Ultrasonic Transducer Enables Bubble-Based Stirring and Reconfigurable Particle Patterning. in *2023 22nd International Conference on Solid-State Sensors, Actuators and Microsystems (Transducers)* 903–906 (2023).
24. Weekers, B. P., Lagae, L., Rottenberg, X. & Rochus, V. ALN-based PMUT Arrays for Dexterous Cell Handling. in *2023 22nd International Conference on Solid-State Sensors, Actuators and Microsystems (Transducers)* 804–807 (2023).
25. Xu, G. et al. Acoustic Characterization of Polydimethylsiloxane for Microscale Acoustofluidics. *Phys. Rev. Appl.* 13, 054069 (2020).
26. Holman, J. B., Zhu, X. & Cheng, H. Piezoelectric micropump with integrated elastomeric check valves: design, performance characterization and primary application for 3D cell culture. *Biomed Microdevices* 25, 5 (2023).
27. Afrasiab, H., Movahhedy, M. & Assempour, A. Proposal of a new design for valveless micropumps. *Scientia Iranica* 18, 1261–1266 (2011).
28. Okamoto, Y., Ryoson, H., Fujimoto, K., Ohba, T. & Mita, Y. On-Chip CMOS-MEMS-Based Electroosmotic Flow Micropump Integrated With High-Voltage Generator. *Journal of Microelectromechanical Systems* 29, 86–94 (2020).

29. Chen, X.-M. et al. A Capillary-Evaporation Micropump for Real-Time Sweat Rate Monitoring with an Electrochemical Sensor. *Micromachines* 10, 457 (2019).
30. You, R. et al. A Miniaturized Wireless Micropump Enabled by Confined Acoustic Streaming. *Research* 7, (2024).
31. Zhang, Y. et al. An integratable acoustic micropump based on the resonance of on-substrate sharp-edge micropillar arrays. *Lab Chip* 25, 2338–2348 (2025).
32. Gao, Y., Wu, M., Lin, Y., Zhao, W. & Xu, J. Acoustic bubble-based bidirectional micropump. *Microfluidics and Nanofluidics* 24, (2020).
33. Mane, S., Behera, A., Hemadri, V., Bhand, S. & Tripathi, S. Micropump integrated white blood cell separation platform for detection of chronic granulomatous disease. *Microchimica Acta* 191, (2024).
34. Akagi, J. et al. Integrated chip-based physiometer for automated fish embryo toxicity biotests in pharmaceutical screening and ecotoxicology. *Cytometry Part A* 85A, (2014).

## Figures

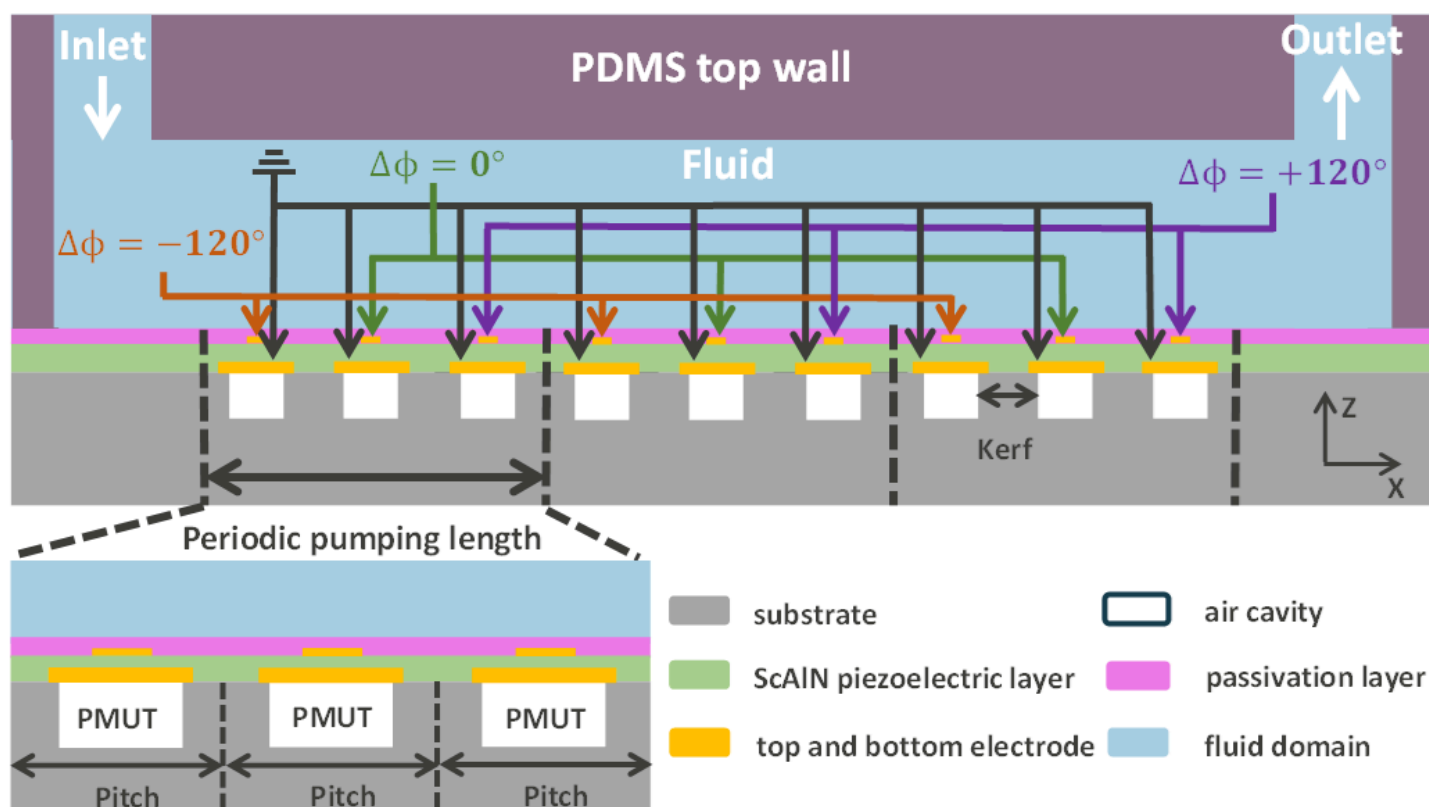


**Figure 1**

Cross section of a rectangular membrane PMUT array micropump. PMUT array (9 x 1 unit cells) was fully embedded inside the polydimethylsiloxane (PDMS) microfluidic channel (height of 360  $\mu\text{m}$ ). Inlet and

outlet reservoirs were designed at the terminals of the microfluidic channel. Each periodic pumping section consists of three PMUT cells.

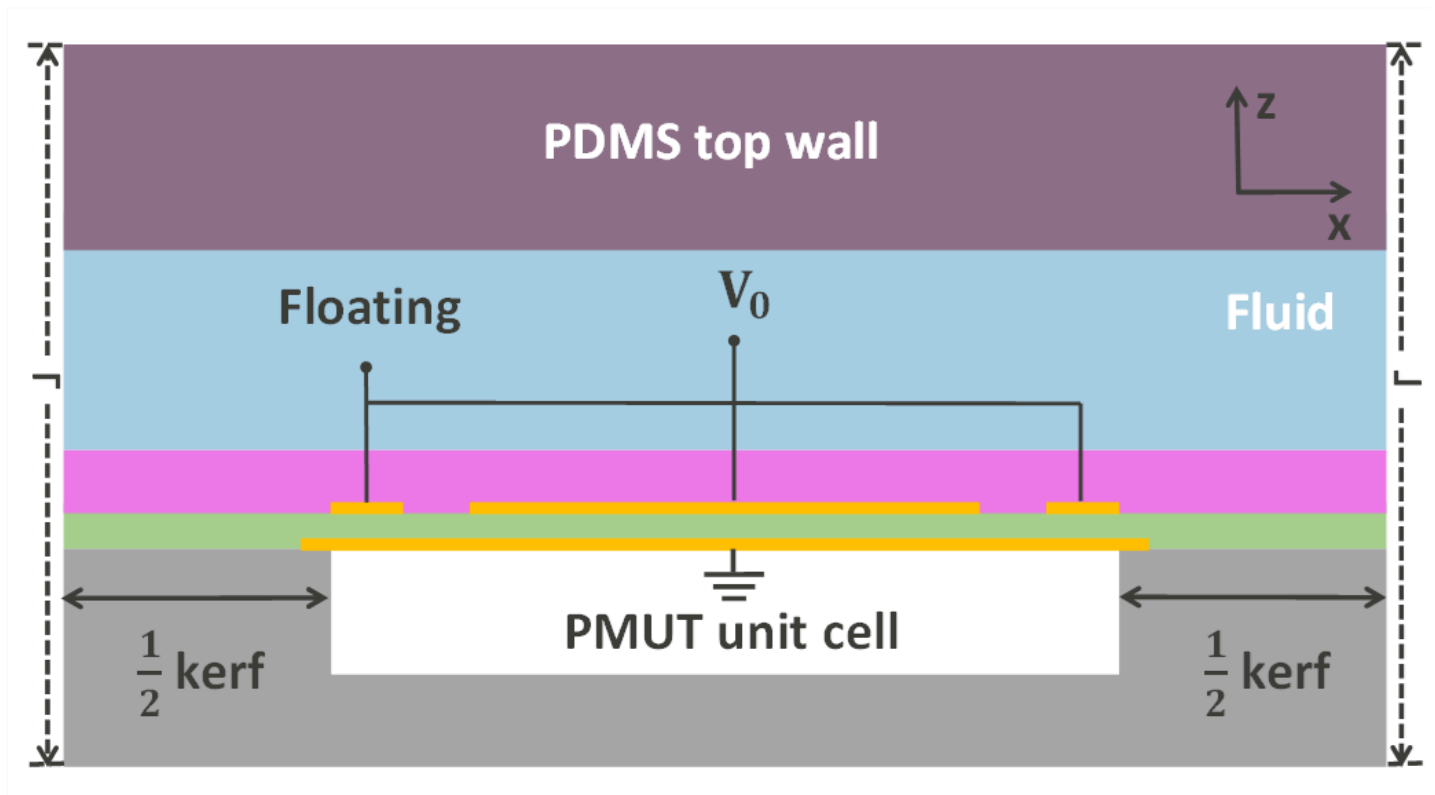


Figure 2

Cross section of one third of the periodic pumping section of the PMUT array micropump, containing one PMUT unit cell. An AC signal is applied to the inner top electrode, the outer electrodes are kept floating, and the bottom electrode is grounded.

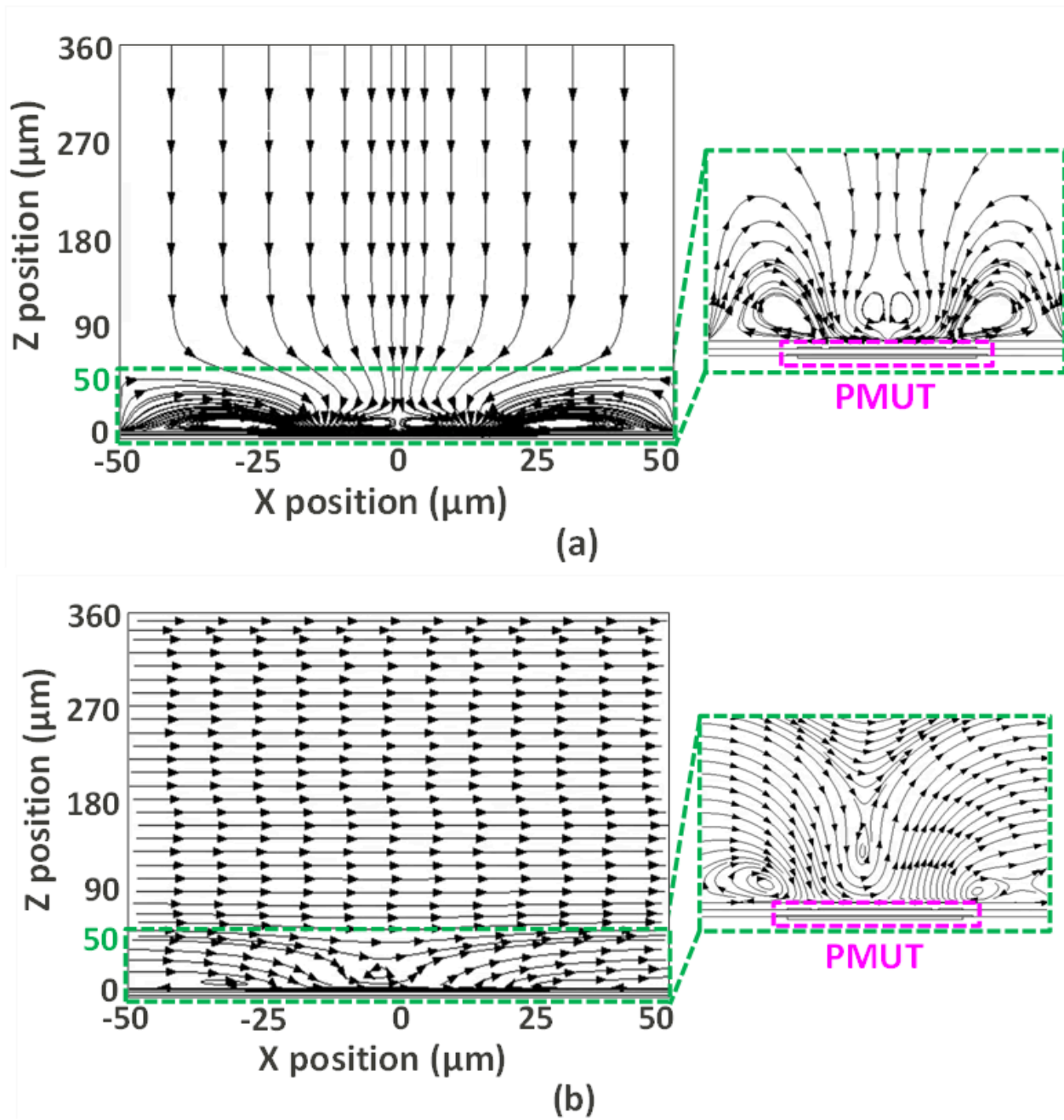


**Figure 3**

Fluid flow streamlines inside the microfluidic channel due to acoustic streaming. Arrows indicate the direction of the flow. (a) Zero net flow when the phase difference between adjacent PMUT cells ( $\Delta\phi$ ) is  $0^\circ$ . (b) Net fluid flow from left to right when the phase difference between adjacent PMUT cells ( $\Delta\phi$ ) is  $120^\circ$ .

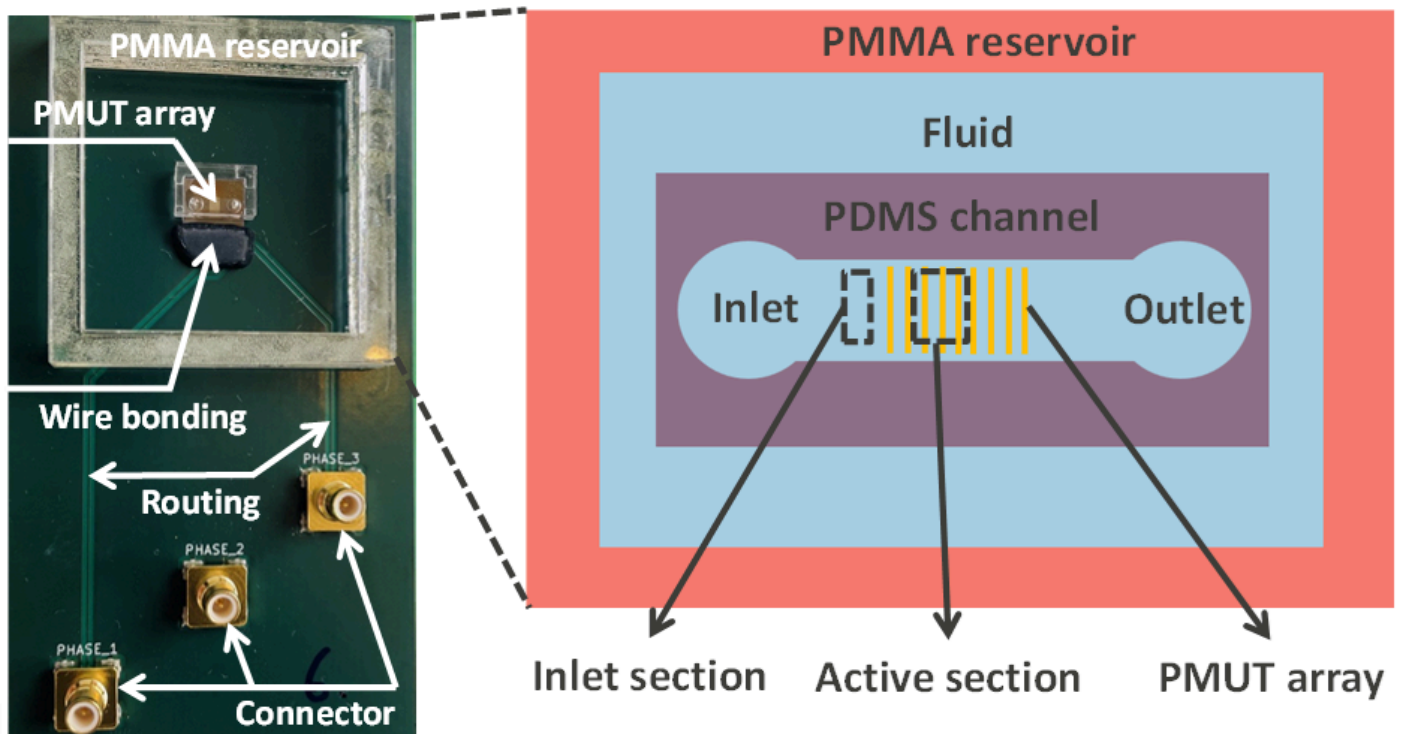


Figure 4

Top view of a rectangular membrane PMUT array micropump. Particles are tracked in both inlet and transducer active sections (observation zones are indicated by dashed rectangles).

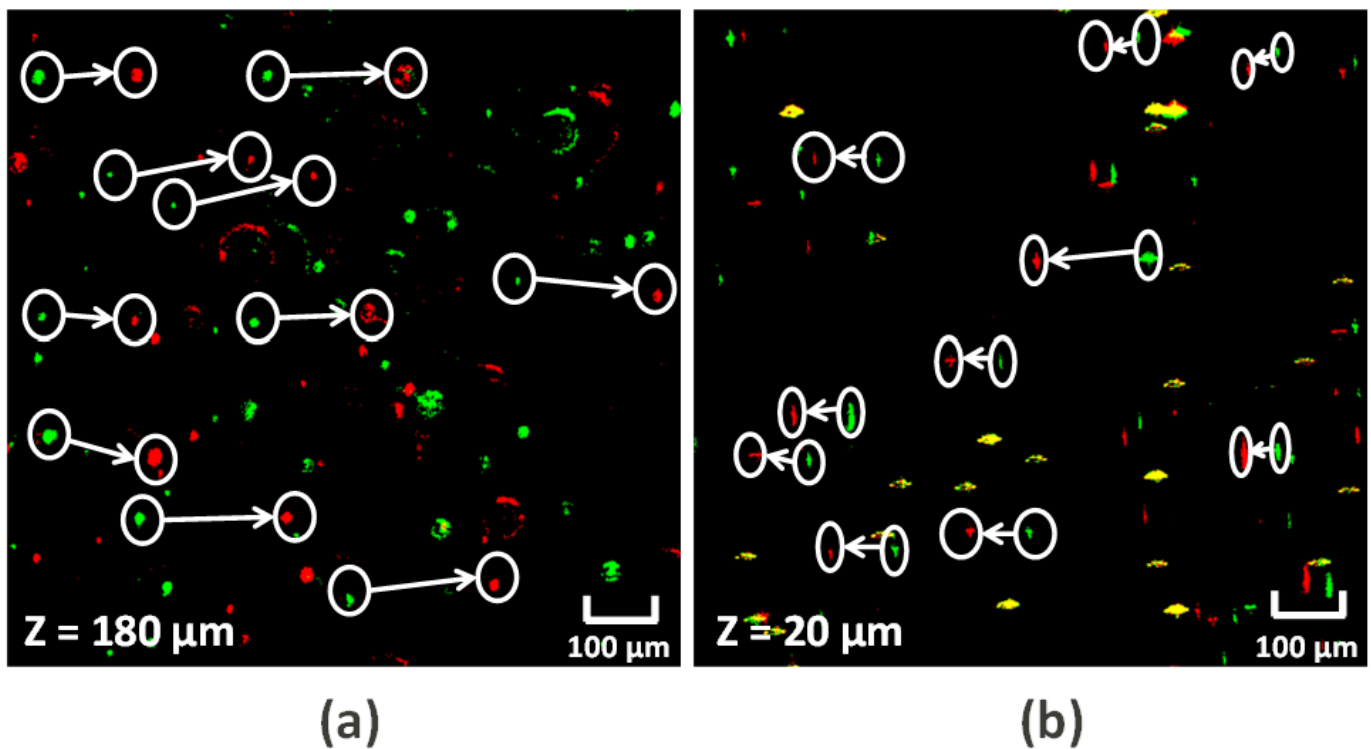


Figure 5

Real time particle tracking in the micropump active section. Green particles represent the initial position, and red particles represent the position after 30 seconds. (a) Particle forward displacement at channel mid-depth ( $z = 180\ \mu\text{m}$  from PMUT surface). (b) Particle reverse displacement near channel bottom ( $z = 20\ \mu\text{m}$  from PMUT surface).

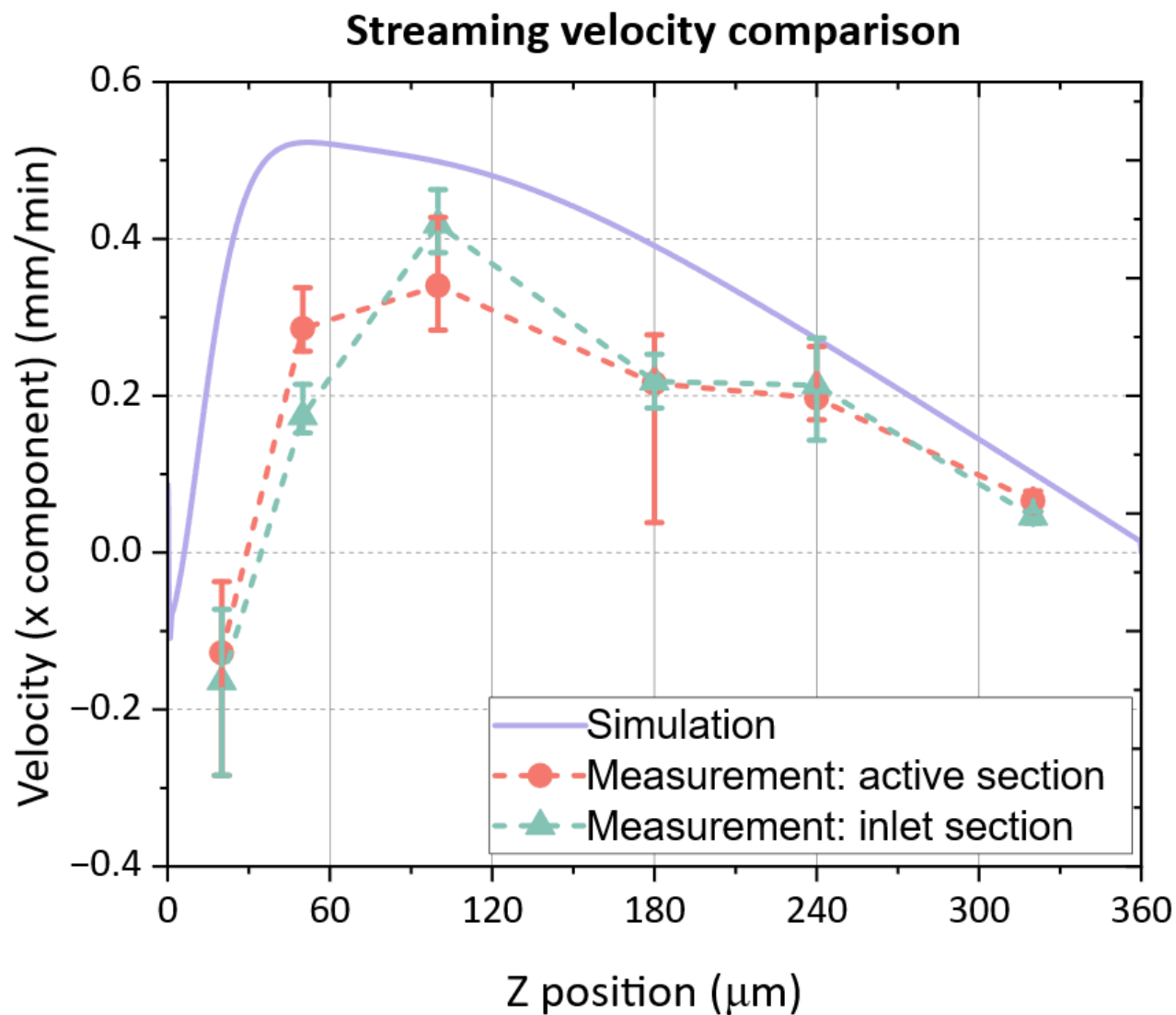


Figure 6

Fluid velocity comparison between simulation and particle tracking measurements for 6 different Z positions along the channel height relative to the PMUT surface. Each PMUT was driven at  $40\ V_{pp}$  with a phase difference of  $120^\circ$  between adjacent PMUTs.

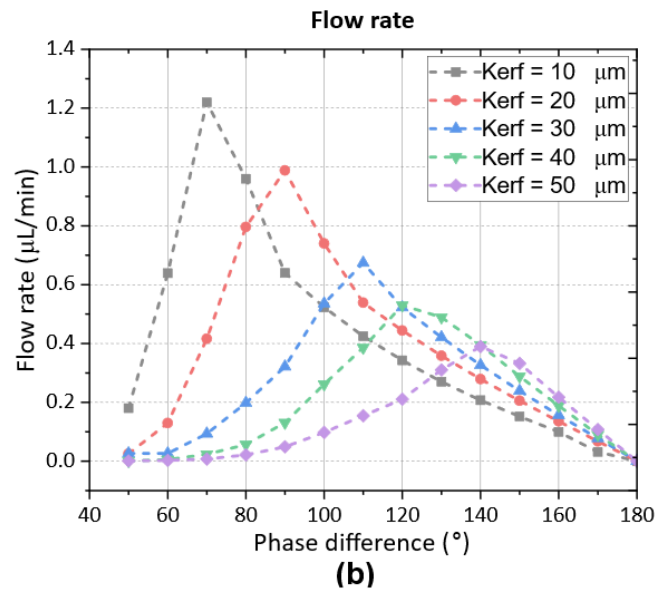
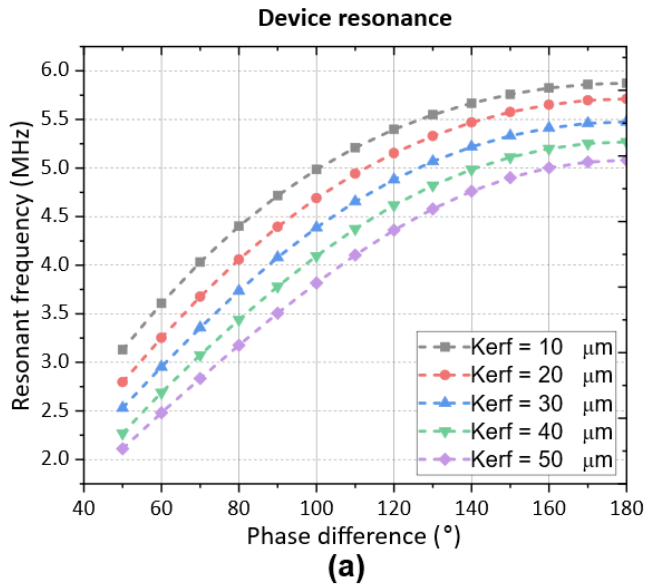


**Figure 7**

Numerical simulation results via parameter sweep for optimization of the micropump's flow rate. Results shown as a function of phase difference for different kerfs: (a) Resonant frequency and (b) Net flow rate produced under resonance.

## Supplementary Files

This is a list of supplementary files associated with this preprint. Click to download.

- [ChenWuSupplementary.docx](#)