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Three-phase ScAlN-based PMUT-driven acoustic streaming micropump

Chen Wu^{1,2✉}, Grim Keulemans², Benjamin Jones², Xavier Rottenberg², Veronique Rochus² and Paul Heremans^{1,2}

Abstract

Scandium-doped aluminum nitride (ScAlN)-based piezoelectric micromachined ultrasonic transducer (PMUT) arrays have attracted increasing attention in acoustofluidics for micro total analysis systems (μ TAS), particularly for applications involving acoustic radiation force for bioparticle manipulation and cell manipulation. 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 driven by a set of AC signals with a 120° phase difference between adjacent PMUT cells to induce directional streaming flow. The device features a compact active area of 1.2 × 1.6 mm and demonstrates a volumetric flow rate of 0.12 μ L/min, in good agreement with predictions from numerical multiphysics simulations. Further numerical optimization suggests that the flow rates of 1.0 μ L/min are achievable by optimizing the array kerf (lateral spacing between adjacent PMUT cells) 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 μ 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 (μ 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 pneumatic¹ and piezoelectric micropumps², 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 separation³ 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 applications⁶. Thermoelectric devices create temperature gradients under applied voltages, which can halt fluid flow or enable on-chip single-cell cryopreservation⁷. Optical tweezers⁸ 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 labeled cells^{9,10}.

Apart from the above-mentioned devices, acoustofluidic devices, which utilize acoustic waves for the 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 diagnostics¹¹. For instance, acoustofluidic centrifugation technique applied for nanoparticle enrichment and separation¹²; harmonic acoustofluidics for dynamic and selective particle manipulation¹³; surface acoustic waves for three-dimensional (3D) manipulation of single cells¹⁴; digital acoustofluidics for contactless and programmable

Correspondence: Chen Wu (chen.wu2@imec.be)

¹Electrical Engineering Department (ESAT), Katholieke Universiteit Leuven (KU Leuven), B-3001 Leuven, Belgium

²Interuniversitair Micro-Electronica Centrum (Imec), Kapeldreef 75, B-3001 Leuven, Belgium

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liquid handling¹⁵. 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 intensity¹⁶. For example, the critical particle diameter is determined to be around 2 μm for a spherical polystyrene particle suspended in water at an acoustic frequency around 2 MHz¹⁷. 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 micro-machined 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 transport¹⁸, pumping¹⁹, 1D or 2D cell patterning²⁰, and particle manipulation^{21,22}. 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 purposes²³.

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 μTAS 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 electrode²⁴. 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 electrodes²⁵. 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)²⁶ and Li et al. (for bubble-based stirring and particle patterning)²⁷. Furthermore, by combining both acoustophoretic and dielectrophoretic forces, Weekers et al. demonstrated dynamic T-cell manipulations using 9.5% Sc-doped AlN PMUT actuators²⁸. Also, monolithic ScAlN-based PMUT-CMOS integration has been demonstrated^{29,30}.

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 μTAS applications. Benchmarking to state-of-the-art micropumps from the literature indicates 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 ($\Delta\phi = 120^\circ$) 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 the 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° phase difference applied between the left and right-hand 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

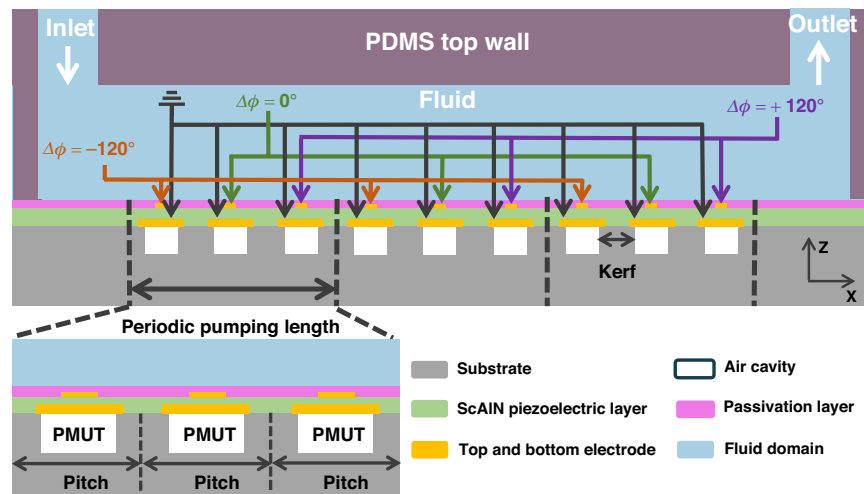


Fig. 1 Cross-section of a rectangular membrane PMUT array micropump. PMUT array (9 × 1 unit cells) was fully embedded in the polydimethylsiloxane (PDMS) microfluidic channel (height of 360 μm). Inlet and outlet reservoirs were designed at the terminals of the microfluidic channel. Each periodic pumping section consists of three PMUT cells

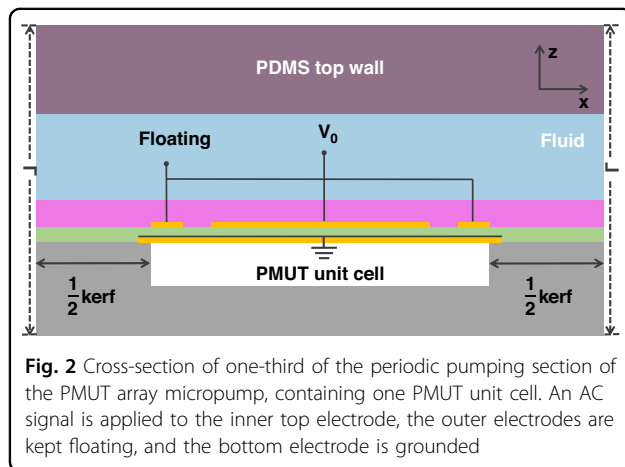


Fig. 2 Cross-section of one-third of the periodic pumping section of the PMUT array micropump, containing one PMUT unit cell. An AC signal is applied to the inner top electrode, the outer electrodes are kept floating, and the bottom electrode is grounded

microfluidic channel (Γ in Fig. 2). The width of the free-standing membrane is 50 μm , and the kerf, which is the distance between adjacent PMUT membranes is 50 μm . The PDMS channel top wall (2 mm thick) is treated as a linear elastic material with a shear viscosity of 3.5 Pa·s, a longitudinal wave velocity of 1030 m/s and a shear wave velocity of 100 m/s³¹. The inner top electrode has a length of 32 μm . The floating outer electrodes (6 μm , at a distance of 3 μm from 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.

A frequency sweep was conducted to extract the resonant frequency of one third of the periodic pumping

Table 1 Dimensions and parameters used in the FEM model

Component	Material	Thickness
Substrate	SiO_2	4 μm
Piezoelectric layer	ScAlN (9.5% Sc doping)	1.3 μm
Passivation	Si_3N_4	1.5 μm
Electrode	Al	0.4 μm
Cavity	Vacuum	0.6 μm
Microfluidic channel	Water	360 μm
Channel top wall	PDMS	2 mm

section, which is 4.37 MHz under water. The maximum out-of-plane displacement of the PMUT membrane is 0.55 nm/ V_{PP} 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 PMUTs driven in-phase ($\Delta\phi = 0^\circ$) and (2) adjacent PMUT driven with a phase difference of $\Delta\phi = 120^\circ$ (see Fig. 3a, b, respectively).

The streamlines are symmetric about the PMUT centreline ($X = 0$) when $\Delta\phi = 0^\circ$, as observed in Fig. 3a; therefore, no net flow. In comparison, by applying a phase difference of 120°, the symmetry is broken, leading to a directional streaming flow shown in Fig. 3b. 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.

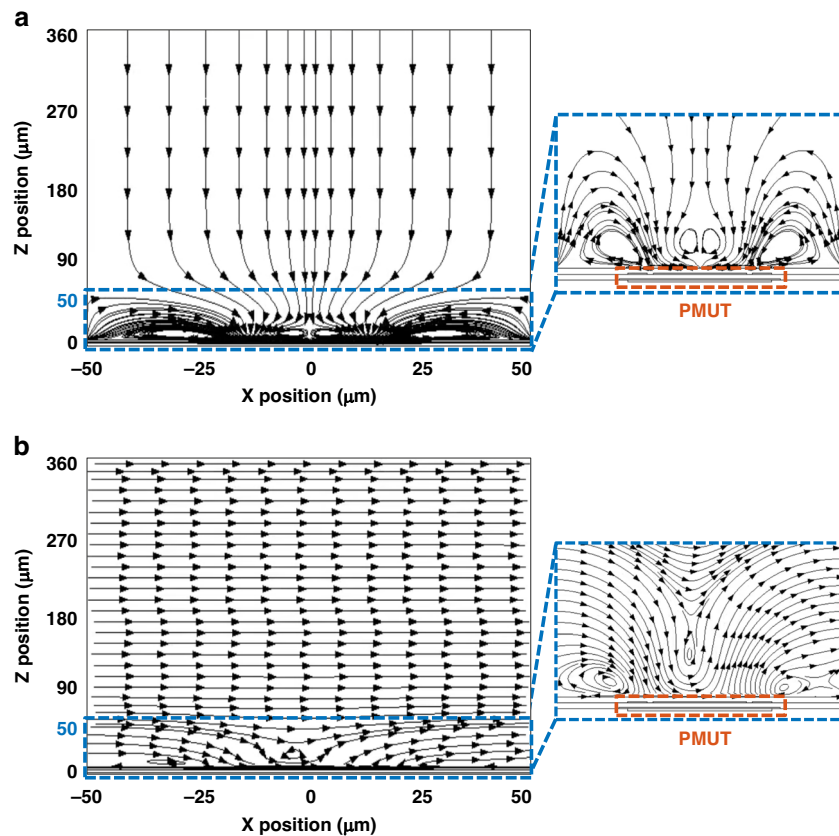


Fig. 3 Fluid flow streamlines inside the microfluidic channel due to acoustic streaming. Arrows indicate the direction of the flow. **a** Zero net flow when the phase difference between adjacent PMUT cells ($\Delta\phi$) is 0° . **b** Net fluid flow from left to right when the phase difference between adjacent PMUT cells ($\Delta\phi$) is 120°

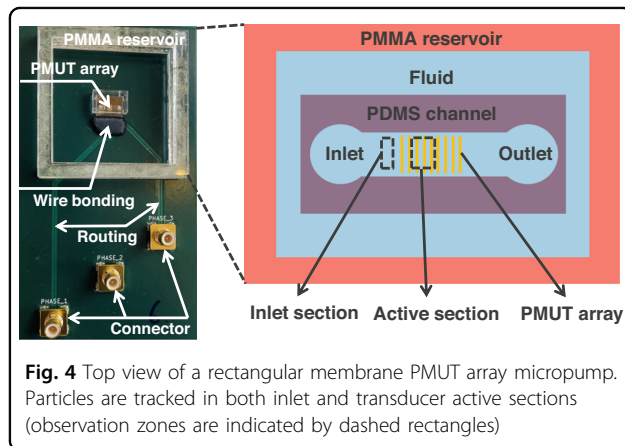


Fig. 4 Top view of a rectangular membrane PMUT array micropump. Particles are tracked in both inlet and transducer active sections (observation zones are indicated by dashed rectangles)

Device description

As shown in Figs. 1 and 4, the PMUT array micropump consists of 9×1 rectangular PMUT unit cells. The thicknesses of different layers are listed in Table 1. The experimental PMUT array had a cell pitch of $100 \mu\text{m}$ and a vibrational membrane width of $50 \mu\text{m}$.

The chip was wire-bonded to a printed circuit board (PCB) for electrical routing. A phase difference of 120° 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 ($\Delta\phi = 120^\circ$) 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 the intended. The maximum out-of-plane displacement magnitude is 0.46 nm/V_{PP} (averaged over 9 PMUTs). The measured displacement phase difference between adjacent cells is $120^\circ \pm 36^\circ$. 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.

Quantification of streaming velocity

Particle tracking measurements followed by μ PIV analysis were conducted to assess fluid flow during actuation of the PMUT-based micropump. The device was driven under 3.92 MHz and 40 V_{PP}, these conditions were used for all the following discussions. Fluorescent particles were imaged within both the PMUT active section and the inlet section (see Fig. 4) at six discrete z-planes spanning the full microfluidic channel height. To highlight characteristic streaming patterns, representative streaming velocity vector fields are shown at $z = 180 \mu\text{m}$, located at mid-depth of the microfluidic channel, and $z = 20 \mu\text{m}$, close to the PMUT array membrane, depicted in Fig. 5a, b. The direction and magnitude of the vectors reveal a forward flow centered in the microfluidic channel, while reverse flow and vortices are observed in the vicinity of the PMUT vibrational membrane. Complete streaming vector fields for all six z-positions are illustrated in Supplementary Fig. 5a–f, confirming a multi-layer acoustic streaming profile which is consistent with numerical simulation results.

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 streaming velocities were quantified using μ PIV technique at six different z positions along the microfluidic channel and compared to simulation results, dashed lines shown in Fig. 6. μ PIV analysis was taken at both the inlet and PMUT array active section (see Fig. 4 for measurement locations) and indicated a maximum flow velocity of 0.33 and 0.35 mm/min at the inlet and active sections, respectively, and a mean streaming velocity of 0.18 mm/min at both the inlet and active section. The simulated streaming velocity is in good agreement with measurements. Vortices are also noted in the μ PIV results, 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 $1.6 \text{ mm} \times 0.36 \text{ mm}$, the resulting volumetric flow rate is approximately 0.12 $\mu\text{L}/\text{min}$, as estimated from the measured streaming velocity profile, with an applied voltage amplitude of 20 V compared to 0.2 $\mu\text{L}/\text{min}$ as simulated. We estimated a basic figure-of-merit by normalizing the measured flow rate to the electrical input power. A total electrical input power of around 5 mW was extracted based on device impedance with a net pumping rate of 0.12 $\mu\text{L}/\text{min}$, the present prototype achieves 8 $\mu\text{L}/\text{min}/\text{W}$ (see Supplementary section 6 Impedance-based estimation of electrical and mechanical power).

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 ($\Delta\phi$). We applied the same FEM model presented above (Fig. 2) that demonstrated a good match in acoustic streaming flow pattern, streaming velocity, and the volumetric flow rate between numerical study and particle tracking experiments. The microfluidic channel height was kept fixed at 360 μm (same as used in previous simulations and experiments). With cyclic periodic boundary conditions applied to both sides, the phase difference $\Delta\phi$ was swept from 50° to 180° with a step of 10° for 5 different fixed values of the kerf, ranging from 10 μm to 50 μm with a step of 10 μm . The resonant frequency of the periodic device and the volumetric flow rates under resonance were evaluated as shown in Fig. 7a, b.

The resonance of the periodic device increases as $\Delta\phi$ 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 result 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 a smaller kerf, while this optimal flow rate occurs at a lower phase difference. The maximum flow rate is identified for an array kerf of 10 μm 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 μm and $\Delta\phi = 120^\circ$). 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.,

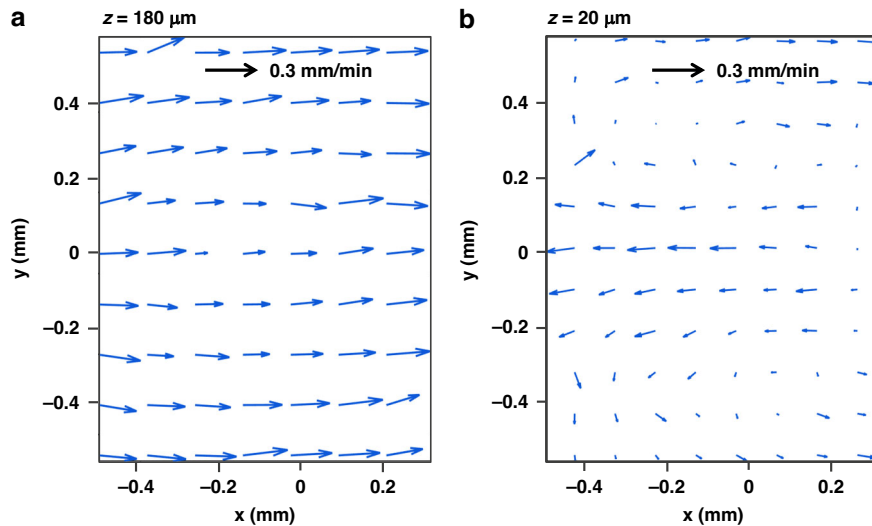


Fig. 5 Streaming velocity vector fields extracted using μ PIV analysis at two representative z -planes within the microfluidic channel. Vectors were obtained from ensemble μ PIV processing of fluorescent particle images in the region spanning both the PMUT active section and the inlet section (see Fig. 4). The coordinate origin corresponds to the geometric center of the PMUT array: $x = 0$ mm marks the beginning of the PMUT row, and $y = 0$ mm aligns with the array's center cutline. A reference arrow (black) indicates a velocity magnitude of 0.3 mm/min. **a** Forward flow at $z = 180 \mu\text{m}$ (mid-depth of the microfluidic channel). **b** Reverse flow and vortices at $z = 20 \mu\text{m}$ (close to the PMUT vibrational membrane)

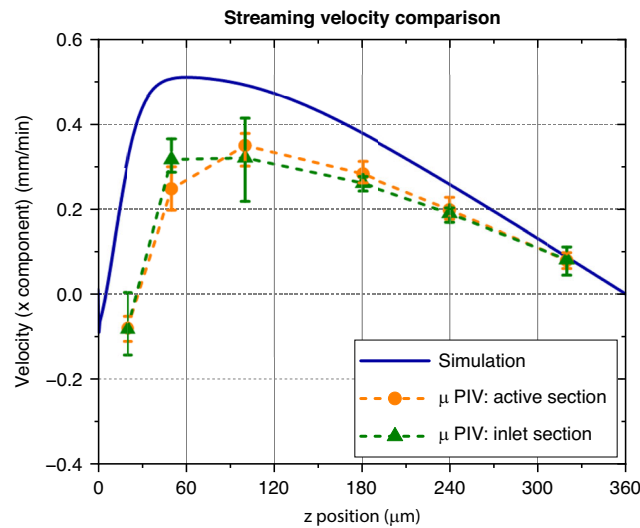


Fig. 6 Comparison of simulated and μ PIV-measured streaming velocities at six z -positions across the microfluidic channel height. PMUT cells were driven at 40 V_{pp} with 120° phase difference. Orange and green markers correspond to μ PIV results from the active and inlet sections (see Fig. 4), respectively. Vertical bars indicate the velocity range within each z -plane, which are defined as the maximum and minimum deviations from the plane-averaged mean values

phase difference of 90°) with an array kerf of 20 μm gives a slightly reduced enhancement in flow rate by a factor of 5 compared to the tested design, while representing 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 V_{pp}), it stands out for its compact footprint and CMOS compatibility.

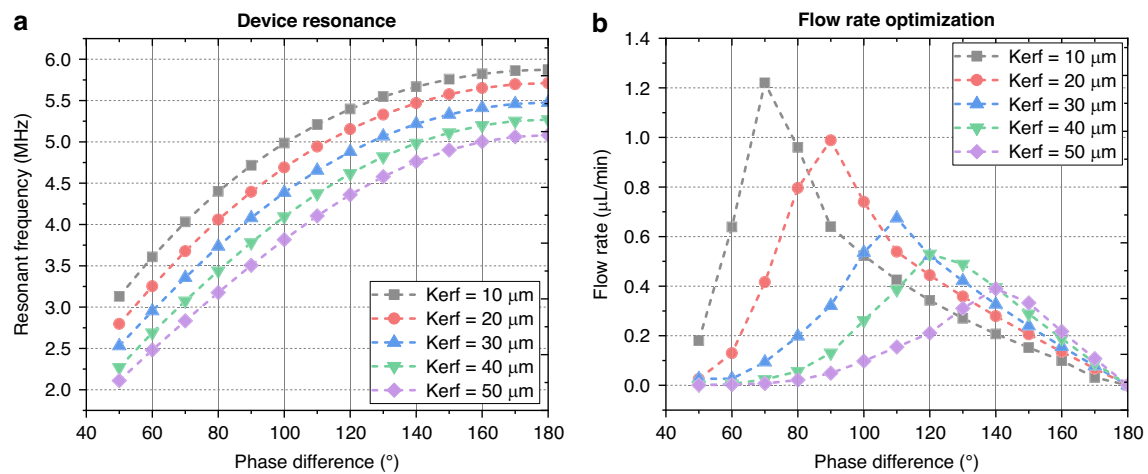


Fig. 7 Numerical simulation results via parameter sweep for optimization of the micropump's flow rate. Results shown as a function of phase difference for different kerfs: **a** Resonant frequency and **b** Net flow rate produced under resonance

Traditional piezoelectric diaphragm micropumps (as an example, see Holman et al.³²) are typically large and rely on PZT, which is not compatible with typical CMOS foundries. Moreover, the typical inclusion of mechanical check valves increases complexity and introduces additional hydraulic resistance. Afrasiab et al.³³ demonstrated a high flow rate generated by a valveless flexural plate wave (FPW) micropump. However, they also use PZT as the piezoelectric material for the membranes, limiting CMOS integration possibilities. Furthermore, the performance of FPW micropumps as described by³² is highly dependent on the coupling between membrane and channel dimensions, which constrains design flexibility. The electro-osmotic micropump developed by Okamoto et al.³⁴ is CMOS compatible; however, it requires high voltages (up to 100 V). Chen et al.³⁵ 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 limit the utility. SAW-driven micropumps, such as the one developed by Wang et al.¹⁹, 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.²³ 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.³⁶) or acoustic air bubbles (Gao et al.³⁷) 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 μ 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 μ 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 systems³⁵ while between 1 and 10 μ L/min are needed in some cell culture chips³³. Applications requiring higher flow rates include blood cell separation platforms³⁸, which can be in the hundred μ L/min range, and wildlife toxicity testing³⁹ requiring up to a thousand μ L/min.

To extend the capability of the micropump to other application domains, improvements in the PMUT membrane displacement are necessary. Increasing the Sc content in ScAlN significantly enhances the piezoelectric response. For example, Zywitzki et al. reported that the piezoelectric coefficient (d_{33}) increases from 8.4 pm/V in undoped AlN to 23.6 pm/V at 33% Sc doping, enabling up to a three-fold increase in actuation efficiency⁴⁰. Similarly, the thickness of the piezoelectric membrane strongly impacts the device performance. We conducted a parametric FEM simulation (Supplementary Fig. 4a, b),

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 ³)	Input voltage or power	Flow rate (μL/min)	CMOS compatibility
Holman et al. ³³	Piezoelectric diaphragm micropump + check valves	12 mm in diameter, 0.2 mm in height	30 – 140 V_{pp} AC	3.4 – 41.8	No (PZT)
Afrasiab et al. ³²	Valveless traveling flexural plate wave micropump	52 × 0.2 × 0.3	40 V_{pp} AC	407.4	No (PZT)
Okamoto et al. ³⁴	Electro-osmotic micropump	7.5 × 0.4 × 0.05	5 V DC	0.167	Yes
Chen et al. ³⁵	Micropores evaporation-driven micropump	Large but not specified, > 10 × 10 × 0.25		0.24	No
Wang et al. ¹⁹	IDT-based SAW micropump	10 × 0.5 × 0.07	7.5 W	0.08	No (LiNbO ₃)
You et al. ²³	GHz Eckart acoustic streaming micropump	9 × 9 × 9	1 W	11800	No (SMR)
Zhang et al. ³⁶	Sharp-edge micropillar-based acoustic streaming micropump induced	25 × 30 × 0.5	54 V_{pp} AC	16.2	No (PZT)
Gao et al. ³⁷	Air bubble-based acoustic streaming micropump	75 × 0.8 × 0.1	7 V_{pp} AC	1.6	No (PZT)
This work	PMUT-based acoustic streaming micropump	3.6 × 1.6 × 0.36	40 V_{pp} AC	0.1 – 1.0	Yes (ScAlN)

demonstrating that decreasing the ScAlN thickness from 1.5 μm to 0.9 μm increases both interface displacement and mean streaming velocity at the channel outlet by a factor of 3 and 1.7, respectively. 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. 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^{41,42}. This dual functionality presents an opportunity for expanded use of this device in μTAS 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 the soft lithography technique. The master mold was 3D printed by Proto Labs Germany GmbH using an ABS-like MicroFineTM 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 centrifuged at 0.3 rcf for 3 min at room temperature, poured over the master, and cured for 2 h 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 h 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, was 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, was 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 v_1 and density ρ_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).

Quantitative acoustic streaming characterization

Micro particle image velocimetry (μPIV) was performed to quantitatively characterize the acoustic streaming velocity field generated by the PMUT micropump. The fluorescent particle displacement was recorded at six different z positions spanning the microfluidic channel height. For each z position, 800 consecutive frames were recorded over 177 s (4.52 fps), covering a sufficiently long duration for ensemble-averaged steady-state streaming vector reconstruction.

Image sequences were analyzed using PIVlab (Version 2.8) in MATLAB, which incorporates state-of-the-art μPIV algorithms validated for microscale, low-speed flows. Velocity extraction followed the workflow presented in Thielicke & Sonntag⁴³ and Raffel et al.⁴⁴ where accurate reconstruction of low-speed microfluidic channel flows has been validated. Prior to cross-correlation, images were pre-processed using intensity normalization and high-pass background subtraction to ensure robust particle feature detection.

Displacement fields were computed using a multi-pass, decreasing-window cross-correlation scheme. Initial interrogation used 64×64 -pixel (px) windows and was refined iteratively to a final window size of 16×16 px with 50% overlap, corresponding to a spatial vector grid of 8 px (approximately 5.2 μm). After each pass, vector fields were shifted to maximize particle pair coherence and suppress spurious vectors, improving accuracy near regions with flow gradients and micro vortices.

Each processed frame yielded approximately 8260 valid vectors within the region of interest (ROI). Based on the particle seeding density (3–10 tracers per 16×16 px window), each frame contained around 4×10^4 tracer particles, which is sufficient to ensure correlation convergence and statistical reliability. Ensemble cross-correlation over the entire 800 frames was applied to maximize signal-to-noise ratio and produce a highly converged mean streaming velocity field.

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Author contributions

C.W.: 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. G.K.: guidance and support for the development of the experimental setup, numerical model, results and analysis and support for the drafting of the manuscript and reviewing/editing. B.J.: guidance and support for the development of the experimental setup, numerical model, results and analysis and support for the drafting of the manuscript and reviewing/editing. V.R.: manuscript reviewing and editing. X.R.: initial device concept, advice supporting various technical and scientific aspects of the work, and manuscript reviewing and editing. P.H.: advice supporting various technical and scientific aspects of the work, and manuscript reviewing and editing.

Conflict of interest

The authors declare no competing interests.

Ethics approval and consent to participate

Not applicable.

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