



A piezoelectric pump with flow-concentrating chamber for multifunctional microfluidic systems

Song Chen¹ · Junjun Zhou¹ · Junjie Li¹ · Zanmei Zou¹ · Zhonghua Zhang¹ · Junwu Kan¹ · Qi Cui² · Tianlong Zhang² · Yigang Shen¹

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Abstract

Piezoelectric pumps have garnered significant interest in such systems owing to their compact form factor, ease of integration, rapid response, and precise fluid delivery. However, simultaneously achieving high output and stable flow remains challenging due to internal flow losses and flow field fluctuations. To overcome this limitation, we propose a piezoelectric pump with a flow-concentrating chamber (PPFC) for multifunctional microfluidic applications. The flow-concentrating chamber creates a low-resistance pathway within the pump chamber, which regulates hydraulic resistance distribution and enhances directional fluid transport. The performance of the PPFC was systematically characterized through numerical simulations, prototype fabrication, and experimental measurements, including on-chip particle tracking velocimetry (PTV). Compared with a standard piezoelectric pump, the PPFC achieved an approximately 30% higher output flow rate and a significant reduction in velocity fluctuation, demonstrating improved pumping efficiency and flow stability. Furthermore, the PPFC was successfully applied to droplet generation and inertial particle focusing, exhibiting performance comparable to that of conventional constant-flow pumping systems. This work provides a simple and scalable strategy for improving the performance of piezoelectric pumps and advancing their integration into multifunctional lab-on-a-chip platforms.

Keywords Piezoelectric pump · Flow-concentrating chamber · Lab-on-a-chip · Droplet generation · Particle focusing

Introduction

Lab-on-a-chip (LOC) systems represent an advanced development of microfluidic technology [1–3] by integrating multiple microfluidic functions, such as particle separation [4–6], sample detection [7, 8], and drug screening [9, 10], into a single platform, thereby enabling the miniaturization and integration [11, 12] of conventional laboratory operations. These integrated platforms provide advantages including high processing speed, efficient sample utilization

[13, 14], and significantly reduced experimental costs [15, 16], enabling broad applications in biomedicine [17], analytical chemistry [18], and environmental monitoring [19]. In LOC systems, precise regulation of fluid behavior is essential because microscale flows are typically dominated by laminar effects, making flow distribution and transport characteristics highly dependent on the design of fluid control components. Therefore, stable and controllable fluid driving is a prerequisite for achieving reliable downstream microfluidic functions.

As the primary fluid-driving components in microfluidic systems, pumps with high integration capability and stable output characteristics play a critical role in advancing microfluidic technologies. The rational selection and design of pumps directly affect the reliability, controllability, and overall performance of microfluidic systems. Pumps in microfluidic systems can be categorized as mechanical or non-mechanical based on their operating mechanisms. Mechanical pumps generate fluid flow through periodic motion of components, such as diaphragm pumps [20] and

Song Chen and Junjun Zhou contributed equally to this work.

✉ Yigang Shen
yigangshen@zjnu.edu.cn

¹ The Institute of Precision Machinery and Smart Structure, College of Engineering, Zhejiang Normal University, Jinhua 321004, China

² College of Mechanical Engineering, Jiangsu University of Science and Technology, Zhenjiang 212100, China

peristaltic pumps [21], offering superior output pressure and control precision. In contrast, non-mechanical pumps rely on physical phenomena such as electric, acoustic, or thermal fields to drive fluid flow; typical examples include electro-osmotic pumps [22] and surface acoustic wave pumps [23]. Compared with non-mechanical pumps, mechanical pumps generally provide higher output pressure and more accurate flow regulation, although their structures are relatively complex. Despite these performance advantages, conventional syringe pumps, a widely used type of mechanical pump, are often bulky and heavy, which severely limits their suitability for portable and point-of-care testing (POCT) applications. On the other hand, many lightweight and miniaturized pumps suffer from insufficient output performance, failing to meet the high-flow-rate demands of applications such as inertial particle focusing and sorting. This trade-off between portability and high output performance has motivated the development of alternative pumping solutions. Among them, piezoelectric pumps have attracted considerable attention for microfluidic applications due to their compact size, ease of integration [24], rapid response [25], and precise fluid delivery capability [26].

Recent studies have focused on enhancing their performance in complex fluid environments. Wang et al. proposed a piezoelectric pump incorporating ramped channels, which effectively reduces flow resistance, enhances inertial effects, and improves output performance via optimized flow path design [27]. Shahrzadi et al. proposed a vane-shaped piezoelectric layer to improve the electromechanical response of the actuator, thereby enhancing the pumping performance and efficiency of the piezoelectric pump [28]. Wang et al. increased the output flow rate by installing airfoil baffles within the pump chamber to regulate the reverse shut-off capacity [29]. Existing research on piezoelectric pumps has made significant progress in performance enhancement, such as output flow rate and output pressure. However, despite these advantages, the practical application of piezoelectric pumps in microfluidic systems remains challenging due to insufficient energy density and flow stability, which are essential for maintaining precise fluid delivery and stable downstream microfluidic functions. Consequently, critical LOC functions that demand both high flow rates and exceptional flow stability, particularly inertial particle focusing, have remained reliant on bulky syringe or constant-flow pumps, limiting system portability and integration.

To address these challenges, we propose a novel piezoelectric pump with a flow-concentrating chamber (PPFC) for efficient and stable fluid manipulation. Different from existing strategies that rely on additional flow-control components or complex geometric modifications, our design introduces a simple yet effective localized chamber-height modulation within the pump chamber. This novel structural

strategy redistributes internal flow resistance and promotes preferential fluid transport, thereby substantially improving pumping performance and flow stability without sacrificing compactness. Compared with conventional piezoelectric pumps, the PPFC demonstrates significantly enhanced output performance and flow stability. As a proof of concept, the PPFC was applied not only to droplet generation but also, more importantly, to inertial particle focusing, a demanding application conventionally driven by syringe or constant-flow pumps. Notably, the PPFC achieved particle focusing performance comparable to that of these conventional systems, successfully bringing this stability-critical function onto a compact piezoelectric platform for the first time. The PPFC enables precise droplet production suitable for digital analysis and high-throughput screening, as well as efficient particle alignment and separation for biological and analytical applications. These results validate the performance advantages of the PPFC and highlight its novelty as an integrated fluid-driving module for LOC platforms, offering a new design-to-application pathway for piezoelectric micropumps in stability-critical and high-flow-rate microfluidic tasks.

Experimental

Microfluidic chip fabrication

The droplet generation chip employed a T-junction channel structure, whereas the inertial particle focusing chip consisted of a straight channel with a flared outlet. The design of the particle focusing chip used in this study was derived from the previous work of our co-author Tianlong Zhang [5], in which the effects of key geometric parameters, including the channel aspect ratio and channel length, on particle separation performance were systematically investigated, and an optimized channel configuration was identified. Based on this established design, the particle-to-channel size ratios were further evaluated for all particles employed in the present study, including polystyrene microbeads (diameters of 1, 3, 5, 7, and 10 μm), *Chlorella* cells (approximately 3–7 μm), and *Escherichia coli* (approximately 1 μm). For the focusing channel with dimensions of 20 μm (width) $\times$ 50 μm (height), the particle-to-channel size ratios were calculated based on the channel height. The corresponding ratios were approximately 0.02–0.20 for polystyrene microbeads, 0.06–0.14 for *Chlorella* cells, and 0.02 for *Escherichia coli*. These ratios fall within the range commonly adopted for inertial particle focusing, providing sufficient hydrodynamic interactions while avoiding excessive particle confinement. Therefore, the selected particle sizes and channel geometry provide

appropriate conditions for achieving effective inertial migration and focusing. Arrays of cylindrical micropillars were incorporated at the chip inlets to remove large impurities. Both microfluidic chips were fabricated using standard photolithography [30] and polydimethylsiloxane (PDMS) soft lithography techniques. Detailed fabrication procedures are provided in Note S1.

Design and fabrication of the PPFC

To broaden the applicability of piezoelectric pumps in microfluidic systems, this study proposes a pump design featuring a flow-concentrating chamber (FC). Figure 1a illustrates the overall assembly of the PPFC. The pump employs a multi-layer stacked design, comprising, from top to bottom: an upper cover, a piezoelectric diaphragm, a sealing ring, a valve cap, a valve body, and a base plate. The upper cover and base plate are fabricated from polymethyl methacrylate (PMMA) via computer numerical control (CNC) machining. The piezoelectric diaphragm consists of a piezoelectric ceramic (PZT-5 H) bonded to a copper substrate. The sealing ring is made of a nitrile rubber O-ring, while the valve assembly is laser-cut from polyethylene terephthalate (PET). Figure 1b details the internal pump structure, which features the pump chamber, FC, valve seats, and threaded holes. The pump chamber is defined as the region enclosed by the piezoelectric diaphragm, the sealing ring, and the base plate. An elliptical FC is formed within the pump chamber and houses the inlet and outlet valve seats, which are used to mount the check valves. These valves are secured using an ultraviolet (UV) light-curing process. All components are finally assembled in a stacked configuration using bolts and nuts to form the complete PPFC. The dimensional parameters of each component are listed in Table 1.

Table 1 Structural dimensional parameters of PPFC

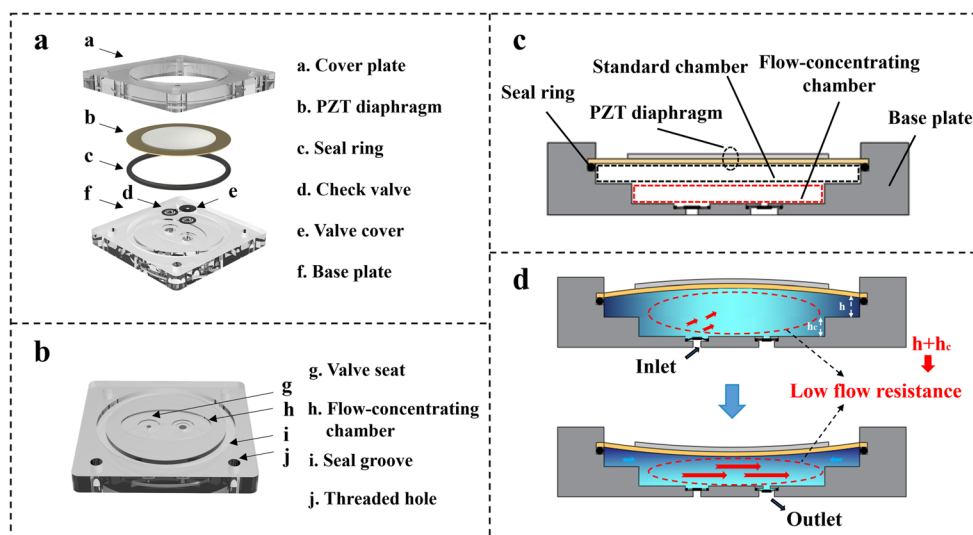
Part	Size/Diameter [mm×mm]/[mm]	Thickness [mm]	Material
Cover plate	45 × 45	5	PMMA
PZT diaphragm	35	0.2	PZT
Sealing ring	41	1.5	Rubber
Check valve	6	0.05	PET
Base plate	45 × 45	5	PMMA

Explanation of the PPFC mechanism

The PPFC is driven by a sinusoidal alternating current (AC) signal, which excites the piezoelectric diaphragm into periodic oscillation. This oscillation induces cyclic pressure variations within the pump chamber, thereby regulating the sequential opening and closing of the inlet and outlet check valves and enabling unidirectional fluid transport. A two-dimensional structural model of the pump is shown in Fig. 1c, while Fig. 1d illustrates the two-dimensional operating principle. During the suction phase, the diaphragm deflects upward, increasing the chamber volume and reducing the internal pressure, which opens the inlet valve and closes the outlet valve to allow fluid inflow. Conversely, during the pumping phase, the diaphragm deflects downward, decreasing the chamber volume and generating a pressure rise that closes the inlet valve and opens the outlet valve, thereby expelling fluid from the chamber. These two phases alternate continuously to achieve sustained pumping operation.

Calculations indicate that the Reynolds number of the fluid flow within the pump chamber is approximately 20. From a macroscopic perspective, the inherently oscillatory flow can therefore be reasonably approximated as a time-averaged steady laminar flow under low-Reynolds-number conditions. In this regime, viscous forces dominate over

Fig. 1 Structure and working principle of PPFC. (a) Exploded diagram of PPFC. (b) Detailed diagram of the base plate. (c) Two-dimensional schematic diagram of PPFC. (d) The working principle of PPFC



inertial effects, and the flow behavior is primarily governed by the balance between pressure gradients and viscous resistance. The governing Navier-Stokes equation reduces to the classical lubrication form:

$$\frac{dp}{dx} = \mu \frac{d^2 u}{dy^2} \quad (1)$$

where p is the pressure, u is the flow velocity, and μ is the dynamic viscosity.

With no-slip boundary conditions:

$$u(0) = 0, u(h) = 0 \quad (2)$$

where h denotes the local channel height.

Solving the equation yields the velocity profile:

$$u(y) = \frac{1}{2\mu} \frac{dp}{dx} (y^2 - hy) \quad (3)$$

The volumetric flow rate per unit width is obtained as:

$$Q = \int_0^h u(y) dy = -\frac{h^3}{12\mu} \frac{dp}{dx} \quad (4)$$

Therefore, the volumetric flow rate scales with the cube of the local channel height under a given pressure gradient, indicating that local geometric variations strongly govern hydraulic conductance and internal flow redistribution. Based on Eq. (4), it can be further inferred that, upon introducing the flow-concentrating (FC) structure, the locally increased chamber height significantly enhances the volumetric flow rate within the FC region, which is ultimately manifested as an improvement in the overall output flow rate of the PPFC.

Furthermore, within this system, the local flow resistance within the chamber can be approximated as:

$$R_h = \frac{12\mu L}{h^3} \quad (5)$$

where L denotes the feature length.

This indicates that there is a strongly non-linear inverse relationship between flow resistance and local chamber height. Consequently, even slight variations in chamber height can result in significant differences in flow resistance.

Moreover, in a multi-region flow system, the flow distribution follows:

$$Q_i = \frac{\Delta p}{R_i} \quad (6)$$

The fluid will preferentially enter areas of low flow resistance, thereby giving rise to a spontaneous flow redistribution process within the chamber. This mechanism, driven by differences in flow resistance caused by the geometric structure, is the fundamental reason why the fluid concentrates along low-resistance paths and enhances the local flow concentration effect.

In short, the FC creates a localized low-resistance pathway that redistributes viscous resistance and establishes a preferential flow toward the outlet, thereby enhancing flow organization and pumping efficiency. By constraining lateral dispersion and suppressing peripheral recirculation zones, the FC yields a more coherent velocity field and reduces flow-rate fluctuations. Collectively, this structural modulation transforms the internal flow from a dispersed transport state to a dominant-pathway-driven regime, simultaneously improving hydraulic efficiency and output stability.

Experimental design

Experimental setup for PPFC structural optimization

To optimize the structure and performance of the PPFC, an integrated simulation-experimental approach was adopted. The geometric parameters of the FC, which serves as the key flow-regulation component of the pump, significantly influence the internal flow organization and overall pumping characteristics. Therefore, three-dimensional fluid-structure interaction (FSI) simulations were first performed in COMSOL Multiphysics using a model consistent with the fabricated prototype. To improve computational efficiency, the mechanical valves were simplified using an equivalent digital valve model that accurately reproduced their rectification behavior. Different FC geometries were investigated by analyzing streamline evolution, velocity distribution, and output performance to identify suitable structural configurations. The optimization process involved two sequential stages: comparison of different chamber shapes to determine the preferred flow-guiding geometry, followed by further optimization of geometric parameters, including aspect ratio and chamber depth. Detailed descriptions of the simulation model, material properties, boundary conditions, and mesh independence validation are provided in Note S2.

Based on the simulation-guided analysis, PPFC prototypes with different chamber geometries and depths were fabricated and experimentally evaluated. The output flow rate and pressure generation capability were measured to validate the simulation results and determine the optimized FC dimensions. Detailed information regarding the experimental setup and measurement instruments is provided in Note S3.

Experimental setup for on-chip flow rate calibration

To quantify the actual flow transport performance of the proposed PPFC and standard piezoelectric pump (SPP) within microfluidic chips, particle tracking velocimetry (PTV) was employed to measure the on-chip flow velocity and calibrate the corresponding flow rate. A schematic illustration of the experimental setup is shown in Fig. 2a. Polystyrene microspheres (30 μm in diameter) were used as tracer particles and prepared as a 0.1 mg/mL suspension. The tracer suspension was driven through a microchannel (width: 200 μm ; height: 100 μm) by both the PPFC and SPP under identical operating conditions. The particle motions were recorded using a high-speed camera (NPX-GS6500UM) at 2000 fps and analyzed frame-by-frame using Kinovea software. Particle velocities were calculated from the displacement and time interval of individual trajectories. The average flow velocity was obtained from multiple representative trajectories, and the corresponding volumetric flow rate was calculated based on the measured velocity and microchannel cross-sectional area.

Experimental setup for droplet generation

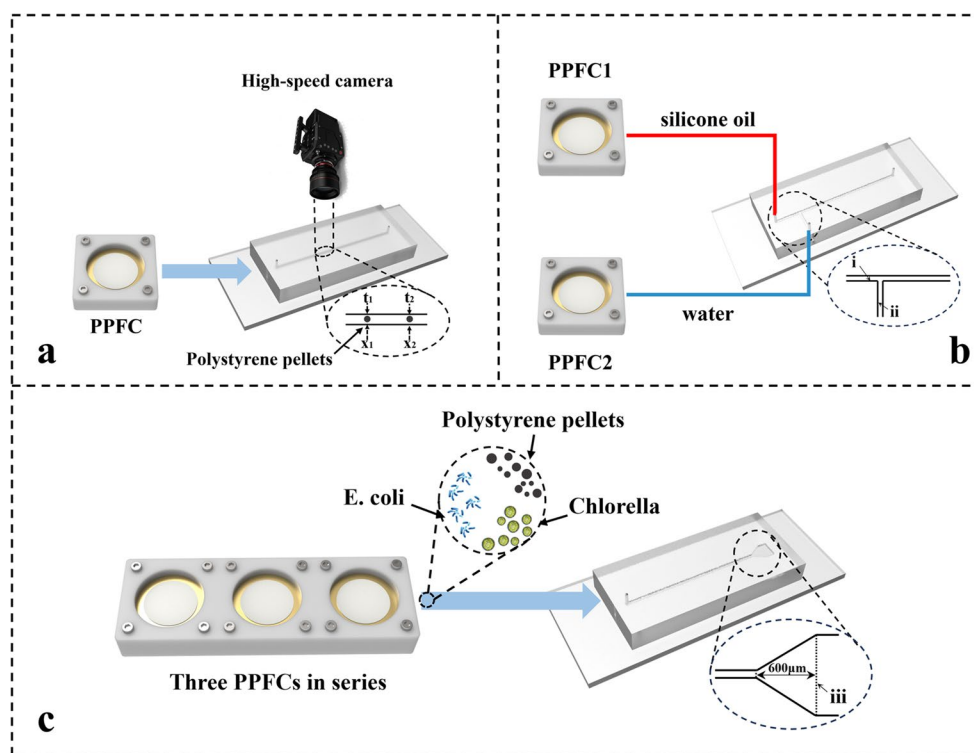
To evaluate the capability of the proposed PPFC for precise microdroplet generation, droplet generation experiments were conducted using a dual-pump-driven microfluidic device. A schematic illustration of the experimental setup

is shown in Fig. 2b. Deionized water and silicone oil were used as the dispersed and continuous phases, respectively, and each phase was independently delivered by an individual PPFC. Based on the voltage-dependent flow regulation capability of the PPFC, the flow rate ratio between the two phases was precisely controlled by adjusting the driving voltage of each pump, enabling active modulation of droplet generation. The microfluidic device consisted of a T-junction geometry with a channel width of 200 μm and height of 100 μm . The continuous phase was introduced through channel i, while the dispersed phase was delivered through channel ii. Droplet formation at the T-junction was monitored in real time using an inverted microscope (XSP-19 C) and recorded using Image View software.

Experimental setup for particle focusing

This experiment aimed to compare the focusing efficacy of the proposed PPFC with that of a constant-flow pump for various particles, thereby validating its application prospects in this field. A schematic diagram of the experimental setup is shown in Fig. 2c. Target particles included polystyrene microspheres (1, 3, 5, 7, and 10 μm in diameter), as well as *Chlorella* cells (approximately 3–7 μm) and *Escherichia coli* (approximately 1 μm). The focusing process utilized viscoelastic fluids, for which solutions of varying concentrations were prepared. For the 3, 5, 7, and 10 μm polystyrene microspheres, a 1000 ppm solution in deionized water was

Fig. 2 Schematic diagram of the experimental setup. (a) Schematic diagram of the on-chip flow rate calibration device. (b) Schematic diagram of a droplet generation device. (c) Schematic diagram of a particle focusing device



used. For the 1 μm polystyrene microspheres, a 3000 ppm solution in deionized water was prepared. For *Chlorella* and *Escherichia coli*, 1000 ppm and 3000 ppm solutions in PBS (phosphate-buffered saline) were used, respectively. Prior to experimentation, all polystyrene microsphere suspensions were ultrasonicated for 10 min to disperse agglomerates. A 0.1% Tween 20 surfactant was added to all particle suspensions to prevent non-specific adsorption and aggregation during flow. During the experiments, the prepared suspensions were injected into a microfluidic chip (channel width: 20 μm ; height: 50 μm). After the flow had fully stabilized, particle focusing at the bell-shaped outlet was observed and recorded using a microscope (HIROX). The captured videos of the focusing process at the outlet were then processed and analyzed using Fiji-ImageJ software. First, time-delayed particle trajectories were overlaid to generate bright-field images of the focusing process. These images were then converted to grayscale, and background noise was reduced. A linear region located 600 μm from the bell-shaped outlet was selected, and its grayscale intensity distribution was extracted. Particle concentration was characterized by grayscale intensity. This intensity distribution was then normalized with respect to both position and intensity values. The focusing efficiency, α , was quantified from the normalized curve and defined as $\alpha = (1 - \text{normalized full width at half maximum}) \times 100\%$.

Results and discussion

FC optimization

Using the established fluid-structure interaction model, simulations were performed for piezoelectric pumps incorporating elliptical and rectangular FCs under identical sinusoidal AC excitation, representing realistic operating conditions. The resulting cumulative flow rates are presented in Fig. 3a. The calculated output flow rates for the two configurations were 18.36 and 18.59 mL/min, respectively, indicating only marginal differences. This suggests that macroscopic output flow rate alone is not sufficiently sensitive to capture the functional variations induced by chamber geometry. Therefore, subsequent analysis focused on the internal flow-field organization.

As shown in Fig. 3b and c, streamline distributions within the pump chamber were extracted from COMSOL simulations. For the rectangular configuration, the overall streamline topology remains generally aligned with the inlet-outlet direction; however, a significant portion of the chamber volume is occupied by weakly engaged regions where streamlines are sparsely distributed. These under-utilized areas are mainly located near the sharp corners and lateral boundary extensions, indicating that part of the geometric space does not effectively contribute to the main transport process.

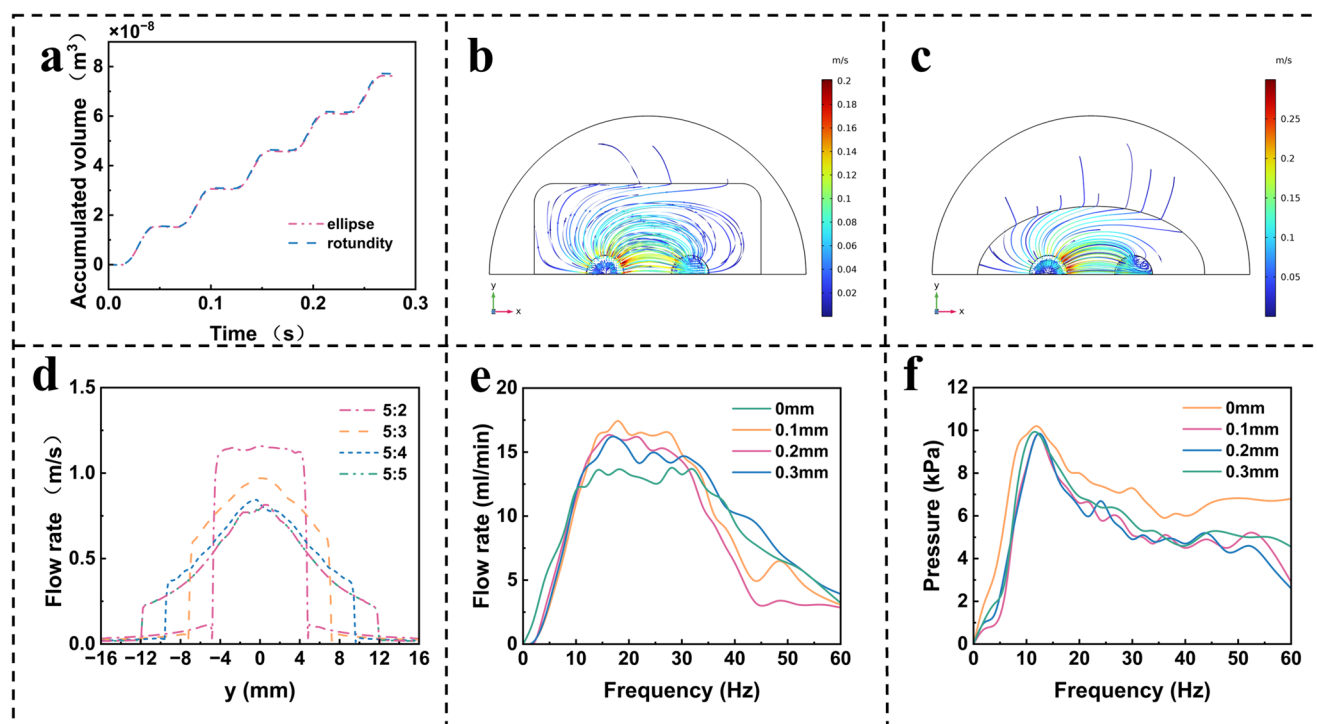


Fig. 3 Simulation analysis and performance testing of PPFC. (a) Simulation of cumulative flow rates of piezoelectric pumps with different flow chamber geometries. (b) Simulation of flow lines in rectangular FC. (c) Simulation of flow lines in elliptical FC. (d) Simulation of flow

velocity in the y axis cross-section of elliptical PPFCs with varying major-to-minor axis ratios. (e) Output flow rate of PPFC. (f) Output pressure of PPFC

This behavior reflects a mismatch between the rectangular boundary enclosure and the naturally concentrated flow pathway, resulting in a relatively high degree of geometric redundancy within the chamber.

To quantitatively evaluate the flow utilization efficiency of different FC geometries, a dimensionless parameter η , defined as the effective flow occupation area ratio, was introduced:

$$\eta = \frac{A_{eff}}{A_{FC}} \times 100\% \quad (7)$$

where A_{eff} represents the effective flow occupation area within the FC region, and A_{FC} represents the total area of the FC. A higher η value indicates that a larger proportion of the designed FC region actively participates in fluid transport. Based on this quantitative evaluation, the calculated η values for the elliptical and rectangular FC configurations were 87.8% and 74.2%, respectively. The higher η value of the elliptical FC indicates more effective utilization of the chamber volume, whereas the rectangular FC exhibits larger ineffective regions caused by abrupt geometric transitions at the corners. The smoothly varying boundary of the elliptical FC provides better geometric-flow matching, enabling a more continuous and compact flow distribution within the chamber. Therefore, compared with the rectangular configuration, the elliptical FC achieves a more efficient flow occupation pattern and was selected as the preferred geometry for subsequent structural optimization.

After identifying the elliptical configuration as the optimal chamber shape, further geometric optimization was conducted by varying the aspect ratio while maintaining the major axis as a fixed constraint. The major axis was set to 24 mm, which is slightly larger than the end-to-end distance between the two valve seats (18 mm). This ensures that both valve seats are fully enclosed within the chamber while providing sufficient geometric allowance to amplify variations in the minor axis, thereby enhancing the sensitivity of the internal flow field to geometric modulation.

Based on the geometric constraint, four elliptical configurations with major-to-minor axis ratios of 5:2, 5:3, 5:4, and 5:5 were constructed. To provide a simplified yet physically consistent evaluation of chamber performance, the assessment criterion was formulated by focusing on the fluid transport capability. In this context, the maximum local velocity within the chamber, V_{max} , is adopted as a representative metric to quantify the intensity of flow transport along the preferential pathway. This parameter reflects the degree of flow concentration and momentum amplification induced by geometric modulation, thereby serving as an effective indicator of the overall transport efficiency of the chamber.

To ensure a fair comparison among different configurations, the velocity metric is normalized by the maximum value observed across all cases:

$$\hat{V}_{max} = \frac{V_{max}}{V_{max}^*} \quad (8)$$

where $(\cdot)^*$ denotes the maximum value within the candidate designs. Consequently, the chamber performance is evaluated using a single unified criterion:

$$\Pi = \hat{V}_{max} \quad (9)$$

Figure 3d presents the simulated results for different aspect ratios, showing the velocity distribution extracted from the y-sectional plane of the chamber. These results provide a direct visualization of how the local velocity field evolves with geometric variation, thereby offering a physical basis for evaluating the flow transport capability of different configurations. The evaluation results show that V_{max} exhibits a clear nonlinear dependence on the aspect ratio, indicating that geometric variation strongly influences the flow acceleration behavior within the chamber. Among all tested configurations, the 5:2 aspect ratio achieves the highest Π , demonstrating the most effective enhancement of flow transport and is therefore identified as the optimal chamber geometry under the current design criterion.

The height of the pump chamber is a critical structural parameter that determines the output performance of a piezoelectric pump. An excessively large chamber height reduces the compression ratio and weakens the pump's self-priming capability, leading to a lower output flow rate. Conversely, an excessively small height, while yielding a higher compression ratio, significantly increases flow resistance due to the confined space, which can adversely affect performance. Therefore, selecting an appropriate chamber height is essential for optimizing pump performance. Based on our previous studies [31], the optimal chamber height for a standard piezoelectric pump to achieve peak output performance is 0.1 mm. Building on this finding, the present study fixed the pump chamber height at 0.1 mm and experimentally investigated the effect of the FC height on pump performance, using output performance as the evaluation metric to determine its optimal value. Experiments were conducted on pump prototypes with FC heights of 0, 0.1, 0.2, and 0.3 mm. The output flow rate and pressure were measured for each configuration, and the results are presented in Fig. 3e and f. The corresponding figure with error bars is provided in Fig. S1 of the Supplementary Material. The experimental results show that the output flow rate initially increased and then decreased with increasing chamber height. At a chamber height of 0.1 mm, the output flow rate reached a peak value of 17.43 mL/min

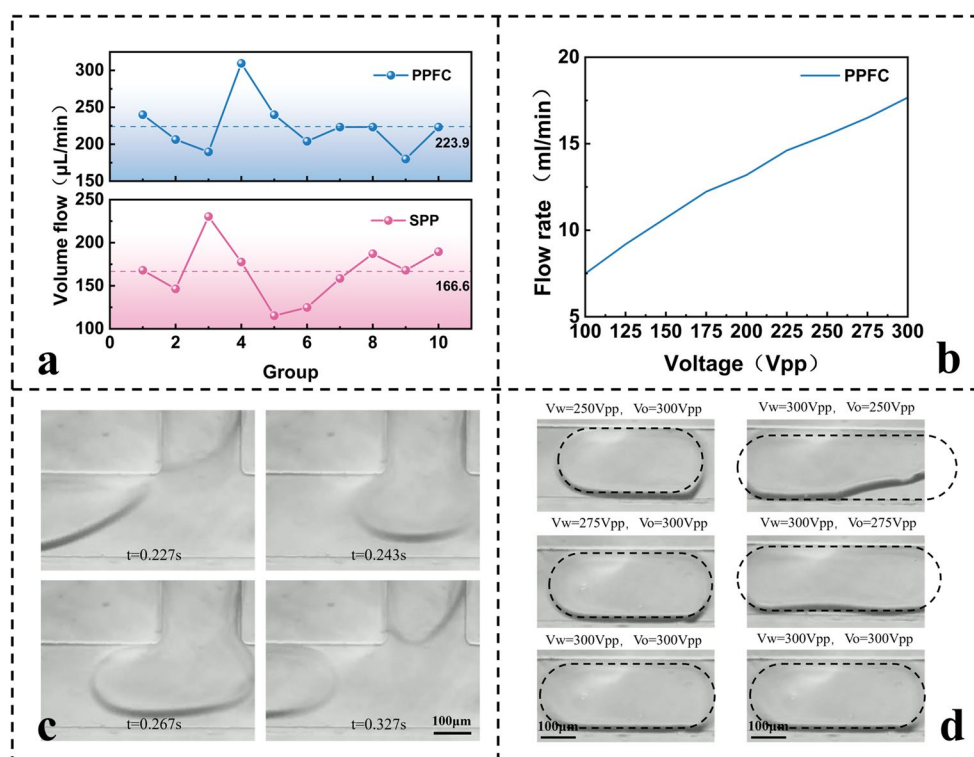
under driving conditions of 300 Vpp and 18 Hz. Regarding output pressure, compared to the control group without a FC (height 0 mm), the introduction of the chamber resulted in a certain decrease. This decrease can be attributed to the increase in total chamber height caused by the FC, which reduces the compression ratio and consequently lowers the output pressure. However, for chamber heights ranging from 0.1 to 0.3 mm, no significant difference in output pressure was observed. This is likely because the FC occupies only a small fraction of the base area; therefore, variations in its height have a limited effect on the total chamber volume, resulting in negligible changes in the compression ratio. In the absence of a FC, a maximum output pressure of 10.2 kPa was achieved at 300 Vpp and 12 Hz. Considering both output flow rate and output pressure, the optimal FC height was determined to be 0.1 mm. Based on the optimized structural parameters, the long-term operational stability of the PPFC was further investigated to evaluate its reliability for continuous microfluidic applications. Specifically, the PPFC was continuously operated for 20 min, during which the output flow rate and pressure were measured at fixed time intervals. The corresponding results are presented in Fig. S2 of the Supplementary Information. Compared with the initial measurements, the final output flow rate and pressure decreased by only approximately 4% and 2.7%, respectively. These results demonstrate that the proposed PPFC maintains stable fluid delivery performance during continuous operation, confirming its excellent operational stability and durability.

On-chip flow rate calibration

Representative particle trajectories obtained from PTV measurements are presented in Fig. S3 and S4. The measured flow velocities were statistically analyzed to evaluate the transport performance of the PPFC and SPP under identical operating conditions (300 Vpp and 18 Hz). The raw data were subjected to statistical screening, and abnormal values caused by transient driving fluctuations were excluded according to the predefined criterion described in Note S4.

As summarized in Fig. 4a, the PPFC exhibited significantly enhanced on-chip transport capability compared with the SPP. The average flow rate increased from 166.56 $\mu\text{L}/\text{min}$ for the SPP to 214.4 $\mu\text{L}/\text{min}$ for the PPFC, corresponding to an approximately 28% improvement. This enhancement is attributed to the flow-concentrating chamber, which reduces local hydraulic resistance and promotes more effective fluid transport within the pump chamber. In addition to the increased average flow rate, the PPFC exhibited improved velocity consistency. The standard deviation of the measured flow velocity decreased from 33.35 for the SPP to 20.95 for the PPFC, indicating reduced velocity fluctuation during operation. The corresponding figure with error bars is provided in Fig. S1 of the Supplementary Material. This improvement suggests that the flow-concentrating chamber not only enhances the transport capacity but also contributes to a more stable flow output, which is highly

Fig. 4 PTV results and droplet generation. **(a)** Comparison of on-chip flow velocities between PPFC and SPP. **(b)** The relationship between the output flow rate of the FCPP and the drive voltage. **(c)** Bright-field image of droplet formation process. **(d)** Droplet length control



desirable for downstream microfluidic applications requiring precise flow regulation, including droplet generation and particle manipulation.

Droplet generation

Since droplet generation is strongly governed by the flow rate ratio between the dispersed and continuous phases, precise regulation of this ratio is essential for controlling droplet dimensions. As demonstrated above, the output flow rate of the PPFC exhibits an approximately linear dependence on the applied driving voltage (Fig. 4b), allowing independent adjustment of the two-phase flow rates by regulating the voltages of the two pumps. To investigate the effect of the two-phase flow rate ratio on droplet formation, numerical simulations were first performed to qualitatively characterize the variation of droplet size under different flow conditions. As shown in Fig. S5, the droplet length exhibited an increasing trend with increasing dispersed-phase flow rate, whereas increasing the continuous-phase flow rate led to a reduction in droplet length. These simulation results reveal the dominant role of the flow rate ratio in regulating droplet dimensions and provide a qualitative understanding of the droplet formation mechanism.

The droplet generation process is illustrated in Fig. 4c, where the aqueous phase was periodically confined and sheared by the continuous oil phase at the T-junction, resulting in the generation of uniform droplets. As shown in Fig. 4d, the droplet length exhibited a clear dependence on the relative flow rates of the two phases. Both PPFCs were operated at a fixed frequency of 18 Hz. The continuous phase pump (PPFC 1) was maintained at 300 Vpp, while the voltage of the dispersed phase pump (PPFC 2) was increased from 250 to 300 Vpp. The resulting increase in dispersed-phase flow rate led to a gradual increase in droplet length. Conversely, when PPFC 2 was fixed at 300 Vpp and the driving voltage of PPFC 1 was increased from 250 to 300 Vpp, the enhanced continuous-phase flow reduced the droplet length. The experimental observations agree well with the simulation results, further confirming that

the PPFC enables effective control of droplet dimensions through precise regulation of the two-phase flow rate ratio.

In addition to droplet size regulation, the uniformity of the generated droplets was quantitatively evaluated by calculating the coefficient of variation (CV) of droplet length, defined as the ratio of the standard deviation to the average droplet length. The calculated CV values were maintained within an acceptable range (5–10%), indicating good droplet size uniformity and stable two-phase flow control using the PPFC. The detailed CV analysis is provided in the Supplementary Material (Fig. S6).

These results demonstrate that the PPFC enables independent and precise regulation of multiphase flow conditions, providing stable control over droplet generation. The capability of actively tuning droplet dimensions highlights the potential of the PPFC as an integrated fluid-driving module for downstream microfluidic applications.

Particle focusing

In microfluidic applications such as focusing, mixing, and separation, the channel dimensions within chips are typically on the micrometer scale, resulting in significantly increased flow resistance. Consequently, the driving devices must provide sufficient output pressure to overcome this high flow resistance. Constant-flow pumps, owing to their design, can achieve high-precision, highly stable flow rates within microfluidic chips, leading to their widespread adoption in this field. However, as microfluidic systems continue to evolve toward miniaturization and integration, the relatively large size of constant-flow pumps has become a primary constraint on their integration into compact systems. Piezoelectric pumps are highly suitable for microfluidic systems due to their compact size, ease of integration, rapid response, and precise fluid delivery capabilities. Furthermore, their design enables them to generate high output pressures when operated in series, effectively overcoming the flow resistance within the chip. This makes them well-suited for use in integrated, miniaturized microfluidic systems. Table 2 summarizes representative piezoelectric microfluidic devices

Table 2 Published parameters relating to piezoelectric pumps

First author and year	Size (mm)	Flow rate (mL/min)	Pressure (kPa)	Optimization Strategy	Advantages	Application scenarios
Wang [29] (2023)	50 × 50 × 20	235.56	0.8	Integrated airfoil baffles	Improved flow rate	Fluid conveyance
Chen [32] (2024)	30 × 30 × 87	197.4	11.0	Integrated flexible support structure	Improved output performance	Fluid conveyance
Fang [33] (2024)	60 × 60 × 9	24.07	9.4	Integrated bluff body	Improved resistance to blockages	Droplet generation
Chen [31] (2025)	45 × 45 × 15	13.24	5	Integrated PTFE	Bubble detection	Bubble sensing
Zhang [34] (2026)	35 × 35 × 25	/	/	Valve seat optimization	Droplet generation and control	Droplet generation
This Paper	45 × 45 × 10	17.43	9.8	Integrated FC	Improved flow rate and stability	Multifunctional microfluidics systems

reported recently. Various aspects, including structural configuration, optimization strategies, output performance, and application scenarios, are compared to provide a comprehensive assessment of the development status and performance enhancement strategies of piezoelectric pumps. Based on this comparative analysis, the advantages and potential of the proposed PPFC can be further understood. To further evaluate the practical potential of the proposed PPFC, a SWOT analysis was conducted, as summarized in Fig. S7. The analysis highlights the technical advantages of the FC-based flow regulation strategy, while also identifying current limitations and future development opportunities for integration into multifunctional LOC systems.

This section further validates the feasibility and practicality of the PPFC for microfluidic applications through particle focusing experiments. The experiment employed three

pumps in a series configuration to achieve particle focusing within a microfluidic chip. The results were compared with the focusing efficacy of a commercial constant-flow pump (CFP). Figure 5a presents an exploded view of the three-pump series configuration used in these experiments. The key performance parameters of the single PPFC and the triple-series PPFC are summarized in Table S1. Compared with the single-pump configuration, the series connection of three PPFCs significantly enhanced the pressure generation capability, enabling the pump system to meet the requirements of microfluidic applications involving increased hydraulic resistance. As shown in Fig. 5b, under driving conditions of 300 V_{pp} and 18 Hz, the three-pump series configuration achieved a maximum output pressure of 51.1 kPa. Under driving conditions of 300 V_{pp} and 36 Hz, the maximum output flow rate reached 25.31 mL/min.

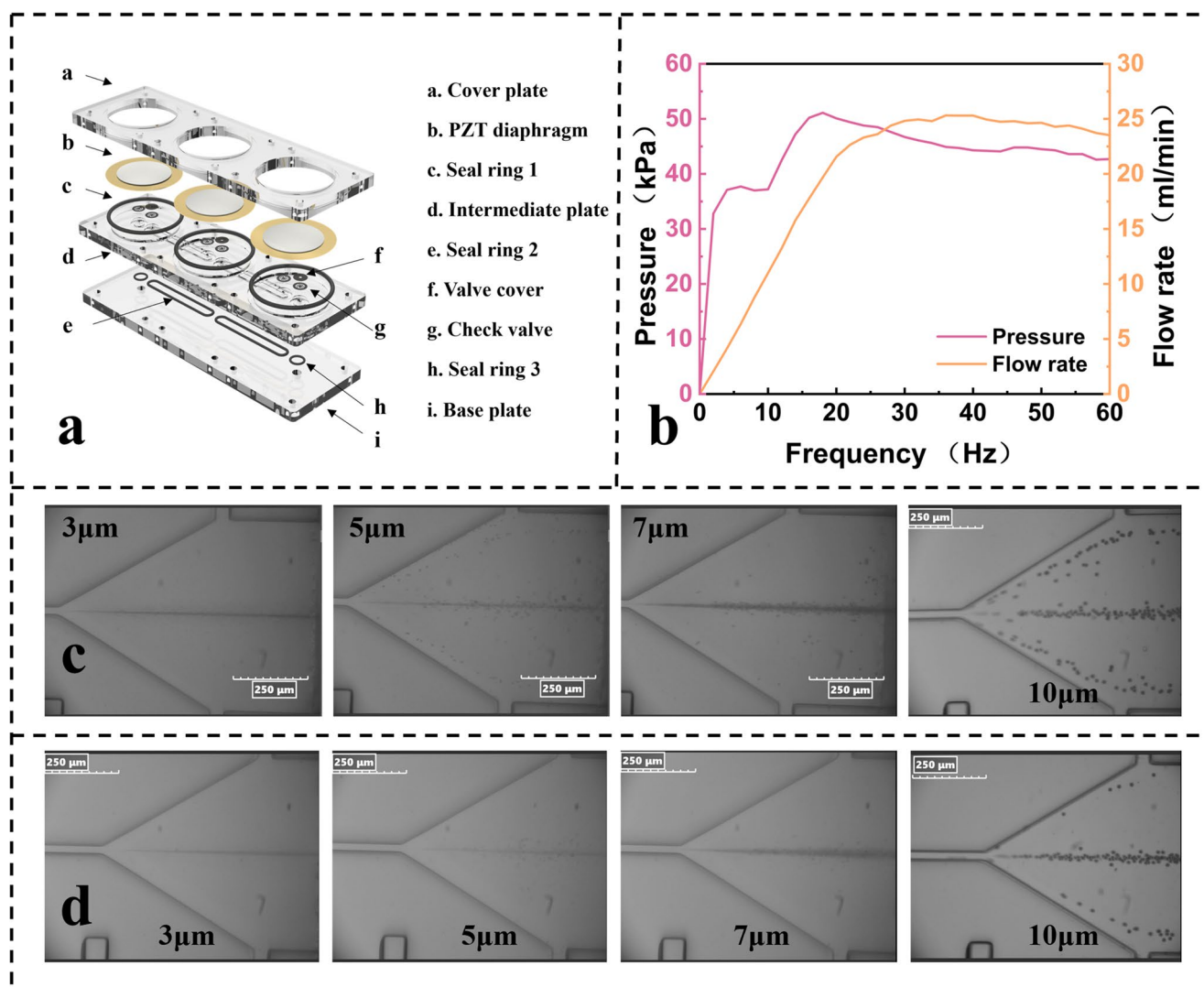


Fig. 5 Particle focusing. (a) Schematic diagram of a triple-series configuration FCPP explosion. (b) Output performance of the FCPP in a triple-series configuration. (c) Overlay plot of focusing time delays

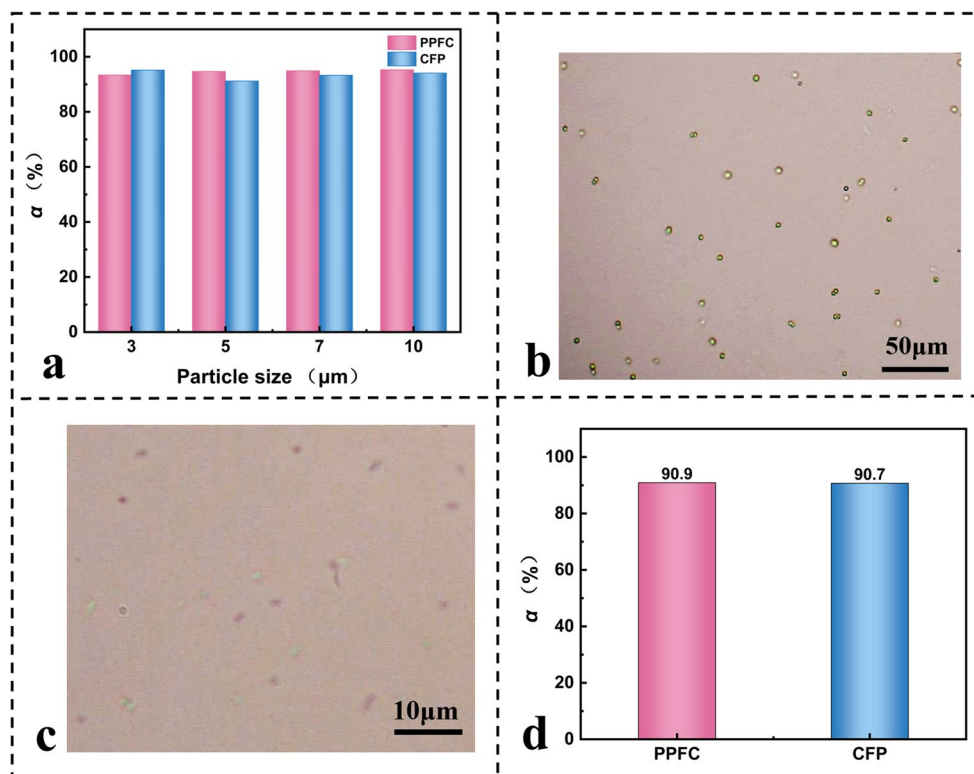
for PPFC with polystyrene microspheres of different diameters. (d) Overlay plot of focusing time delays for CFP with polystyrene microspheres of different diameters

To assist focusing, a 1000 ppm viscoelastic solution was used. Both the proposed PPFC (driven at 300 Vpp, 36 Hz) and the CFP (set to 10 $\mu\text{L}/\text{min}$) were employed to drive target particles of various sizes. The recorded focusing videos were processed using time-delay superimposition in ImageJ, yielding the particle trajectory images shown in Fig. 5c and d. The results indicate that the focusing efficacy achieved with the two driving methods was comparable. Distinct focusing fringes were observed for the 3, 5, 7, and 10 μm particles. However, for the 1 μm particles, focusing performance was suboptimal under both conditions. Corresponding videos are provided in Video S1 and Video S2. This suboptimal performance may be attributed to the reduced elastic lift acting on the 1 μm particles, combined with their pronounced Brownian motion, which hinders stable focusing. To enhance the elastic lift, a 3000 ppm PEO (Polyethylene Oxide) solution was used to assist in focusing the 1 μm particles. The experimental results show that, compared to the 1000 ppm concentration, the 1 μm particles no longer dispersed completely at the outlet but formed a narrow band, demonstrating a certain degree of focusing effect. The corresponding videos are provided in Video S3 and Video S4. The normalized grayscale curves for the 3, 5, 7, and 10 μm particles are provided in Fig. S3. The corresponding focusing efficiency, α , for each particle size is shown in Fig. 6a. For all particle sizes, α exceeded 90%, indicating excellent

focusing performance under the experimental conditions. Furthermore, the focusing efficiency achieved with the PPFC closely matched that of the CFP, further validating the pump's feasibility and reliability for this application. This result highlights its strong potential for use in microfluidic particle focusing applications.

The focusing performance was further evaluated using two types of biological particles with different sizes: *Escherichia coli* (approximately 1 μm) and *Chlorella* (3–7 μm). Microscopic images of the two particles are shown in Fig. 6b and c. For *Escherichia coli*, a 3000 ppm PEO solution was used to assist focusing. Both the proposed PPFC (driven at 200 Vpp, 36 Hz) and the CFP (set to 3 $\mu\text{L}/\text{min}$) were employed. Corresponding videos are provided in Video S5 and Video S6. The results show that, for *Escherichia coli*, both pumps produced a narrow beam along the central axis in a 3000 ppm PEO solution. This indicates that, at this concentration, the particles experience sufficient elastic lift, while the lower flow rate allows for a longer equilibration time, enabling effective focusing even for submicron-sized biological particles. For *Chlorella*, a 1000 ppm PEO solution was used to assist focusing. Both the PPFC (driven at 300 Vpp, 36 Hz) and a CFP (set to 5 $\mu\text{L}/\text{min}$) were employed. Corresponding videos are provided in Video S7 and Video S8. To further visualize the focusing performance of *Chlorella* cells and support the quantitative evaluation, additional representative focusing images and

Fig. 6 Comparison of the focusing effects of PPFC and CFP. (a) Focusing effects of PPFC and CFP on polystyrene particles of different diameters. (b) Microscopic image of *Chlorella*. (c) Microscopic image of *Escherichia coli*. (d) Comparison of the focusing effects of PPFC and CFP on *Chlorella*



the corresponding normalized grayscale intensity profiles were provided in the Supplementary Material (Fig. S9). The intensity profiles were used to calculate the focusing efficiency by quantitatively characterizing the particle distribution across the channel width. The focusing results are shown in Fig. 6d. Both pump types achieved a focusing efficiency of over 90% for *Chlorella*, further demonstrating the versatility and effectiveness of the proposed PPFC for microfluidic focusing applications.

Conclusions

This study proposes a piezoelectric pump incorporating a flow-concentrating chamber for microfluidic applications. The FC establishes a localized low-flow-resistance pathway within the pump chamber, promoting fluid concentration within the FC region and guiding transport toward the outlet. This mechanism reduces flow dispersion and enhances global flow organization, thereby improving both pumping efficiency and flow stability. Through combined numerical simulation and experimental validation, the proposed PPFC demonstrates strong feasibility and applicability in microfluidic systems.

In on-chip flow rate calibration, the PPFC achieves an approximately 28% increase in outlet flow rate compared with SPP. More importantly, the velocity standard deviation decreases from 33.35 (SPP) to 20.95 (PPFC), indicating a substantial reduction in velocity dispersion within the microchannel. This reduction in statistical spread suggests that the PPFC not only enhances overall transport capacity but also effectively suppresses localized velocity fluctuations, resulting in a more homogeneous and stable flow field. In droplet generation experiments, a dual-pump configuration enables stable operation by regulating the flow rate ratio through independent voltage control of each pump, producing predictable droplet generation behavior consistent with established scaling laws. In particle focusing experiments, a multi-pump series configuration is employed to enhance driving capability and overcome high hydraulic resistance in microfluidic channels. This enables efficient inertial focusing of polystyrene microspheres (1, 3, 5, 7, and 10 μm in diameter), as well as *Chlorella* and *Escherichia coli*, with performance comparable to conventional constant-flow pumping systems. These results confirm the adaptability of the PPFC across different microfluidic operation modes.

Overall, the proposed FC-based design leverages the inherent advantages of piezoelectric pumps, including compact structure and fast dynamic response, while simultaneously improving both flow magnitude and flow stability. The

flexible operation modes and stable output further enhance its suitability for integrated microfluidic platforms. With increasing demand for highly integrated LOC systems, the proposed strategy provides a promising route toward structurally compact, functionally versatile, and easily integrable microfluidic pumping solutions.

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Data availability No datasets were generated or analysed during the current study.

Declarations

Generative AI and AI-assisted technologies in the writing process During the preparation of this work the author(s) used ChatGPT in order to improve language and readability. After using this tool/service, the author(s) reviewed and edited the content as needed and take(s) full responsibility for the content of the publication.

Competing interests The authors declare no competing interests.

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