

Simulation and experiments of microelectrode geometry effects on dielectrophoretic particle collection in microfluidic devices

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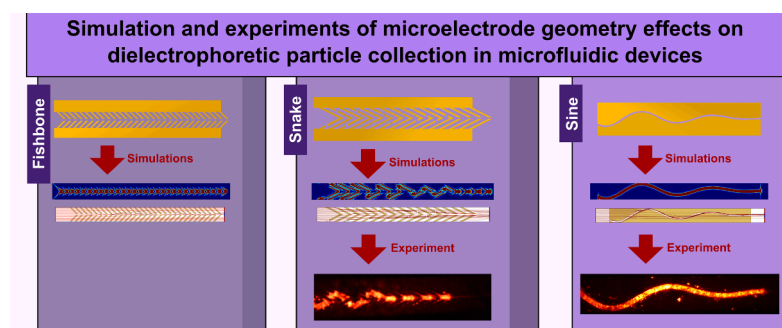
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HIGHLIGHTS

- Design and fabrication of custom-made electrodes and microfluidic channels.
- COMSOL simulations for the dielectrophoretic (DEP) force range and particle tracing.
- Demonstration experiments with 1 μm and 0.1 μm microspheres.

GRAPHICAL ABSTRACT



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ABSTRACT

In a non-uniform electric field, particles show dielectrophoretic (DEP) behavior. This is a phenomenon where particles are attracted or repelled by the electric field depending on their electrical properties, size, the surrounding medium, and the frequency of the applied field. These effects are called positive or negative dielectrophoresis, corresponding to attraction toward or repulsion from the electrodes, respectively. With the right frequency, the particles can be manipulated, separated, or collected, based on their characteristics. We designed and simulated three types of electrodes: "fishbone," "snake," and "sine" structures, and integrated them into microfluidic channels. The electrodes were designed to direct the micron/submicron-sized particles by positive DEP force to the center of the channel, enabling their specific collection and concentration at the outlet. For comparison, we calculated the DEP force for each electrode design, and simulated the particle trajectories within the device. Additionally, we fabricated these devices and experimentally measured their particle collecting efficiency.

1. Introduction

Unlike electrophoresis which acts on charged particles, and normally

works in a uniform electric field, dielectrophoresis (DEP) acts also on uncharged (dielectric) particles, and operates in a non-uniform electric field. The particles show dielectrophoretic activity depending on their

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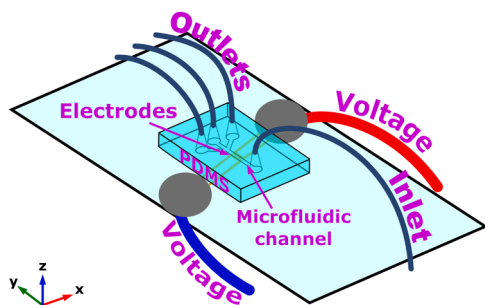


Fig. 1. Design of the device. The microfluidic channel has one inlet and three outlets. The electrodes are placed into the middle of the channel. The channel is made from PDMS and the electrodes are made from gold.

electrical properties, size, the medium and the frequency of the electric field. There are two types of DEP: when the electric field attracts the particles the effect is called positive DEP (pDEP), while when the electric field repels the particles it is called negative DEP (nDEP). The size and direction of the DEP force (F_{DEP}) depends on the complex permittivities of the particle and the medium, according to the following equation:

$$F_{DEP} = 2\pi r^3 \epsilon_0 \text{Re}(\epsilon_m) \text{Re} \left\{ \frac{\epsilon_p - \epsilon_m}{\epsilon_p + 2\epsilon_m} \right\} \nabla |E|^2 \quad (1)$$

Here, ϵ_m and ϵ_p are the complex relative permittivities of the medium (H_2O) and the particle, respectively, ϵ_0 is the vacuum permittivity, r is the radius of the particle and $\nabla |E|$ is the gradient of the electric field. The expression between the curly brackets is called the complex *Clausius-Mossotti* factor. Depending on the sign of its real part, pDEP or nDEP effects can be observed. With the right parameters, we can collect, trap or separate different types of particles from each other [1].

Microfluidic channels can precisely control small fluidic samples. Their volume is about a few μL or even less. Since biological samples are often only available in small quantities, these channels are perfect for their measurements. Also, it is easily combined with the DEP electrode system.

The dielectrophoretic devices can be used for various assays with biological objects, from cells to bio-nanoparticles. The separation and collection of particles can be useful for the characterisation of specific cells or cell components. The traditional separation methods are often expensive, complex and time-consuming, while the microfluidic DEP system's huge advantage is that it is portable, cheap, simple and fast. In biology, the primary use of this system encompasses the identification and separation of malignant, [2–4], dead [5] or bacterial cells [6], and also bio-macromolecules, [7] or cancer biomarkers [8,9]. Because a DEP system is combined with the microfluidic channels, one of its main research areas is the analysis of blood samples. For example, the main aspect of cell-cell separation is the collection of cancer cells from blood. The separation can occur either by trapping [2] or by directing the healthy and the cancerous cells into different outlets (dynamic collection) [3]. Blood cells are also separable from biomolecules, like proteins [7], which facilitates the pretreatment of biosamples. Different diseases accompany different biomarkers, which gives rise to their early detection. Notable examples are the cell-free circulating DNA, which is an early sign of cancer [8], or other biomarkers like extracellular vesicles, which are carrying cancer-related DNA, RNA and proteins [9]. Differently shaped or sized proteins can also be separated from each other [10]. Nevertheless, DEP can be used not just for blood samples, but for other biological assays, as well, such as for enrichment of yeast or *E. coli* cells [11], or for the separation of *E. coli* cells from tap water [12], etc.

Separation of the different biosamples is usually achieved by fine-tuning the voltage and frequency, but the shape of the electrode system is also an important consideration. Many different types of electrode systems are designed for DEP-based separation devices, from the simple to complex. For example, polystyrene microbeads of different sizes are

easily separated from each other with only two electrodes that are at an angle to each other. As the distance between the electrodes increases, the DEP force is decreasing for the same size of object, therefore, it can be used to deflect increasingly larger particles proportional to the distance [13]. Of course, biological samples are also separable with similar simple design. Zhang et al. separated macrophages from breast cancer cells with only two parallel electrodes, by deflecting the macrophage to a side channel with DEP force, while the cancer cells path was not disturbed [14]. But also more complex electrode structures exist. The most common design is the series of parallel electrodes, which are inclined to the channel [4,6,15]. This is a similar design to the previous one, just the more electrodes have greater and gradual effect. There are also circular electrodes which are designed to trap the bioparticles [8, 10]. The DEP force can be enhanced by 3D electrodes by increasing the volume of the particle collection. Martinez-Duarte et al. designed a 3D carbon electrode system, where carbon column electrodes separated the particles [16]. This design can be used in many different ways, such as separation of monocytes from macrophages [17] or collection of lambda-DNA [18]. Another interesting example of application is the combination of wavy microchannels with C-PDMS electrode sidewalls [19].

Our aim is to develop a microfluidic device capable of collecting and separating micron- and submicron-sized particles using pDEP. While most DEP-based collectors and separators are designed for cells, sub-micron structures such as cell organelles also play a crucial role in medical diagnostics. In this size range the particle collection usually happens by trapping these particles inside the device, but if we want to use these particles for other follow-up analysis then their collection in an outlet is necessary. In the following, we use the term “capturing” for a consecutive action of trapping and collection of particles. Although some systems use high voltages to improve performance, such conditions can damage biological particles, making low-voltage operation essential. Also the simple 2D electrodes are important for the easy and fast production of these devices.

To address these needs, we designed a device capable of continuously gathering particles, and directing them into one of its outlets, which is supporting their further biochemical analyses. The system uses low voltage for the collection and has simple 2D electrode design. In a previous study, we successfully applied a special fishbone-like electrode geometry for the dynamic collection of bacterial cells [20]. Now we have used this electrode design as a starting point, with the aim of improving collection efficiency by applying alternative geometries for μm or smaller particles' collection. The final device configuration is shown in Fig. 1.

2. Materials and methods

The particle collection system was based on a lab-on-a-chip platform. It contained an electrode system for collecting particles, and a microfluidic channel to facilitate the flow of the particle-containing fluid. The setup was further supported by a syringe pump and a function generator. The electrodes were designed for positive DEP. The process of making the electrodes and the microfluidic channels was based on a previous study of our group [20].

2.1. The electrode

In the initial step, an S1818 photoresist (S1818 G2 positive photoresist, Microposit) layer was deposited onto a microscope cover glass, using a spincoater (5 sec ramp, 1000 RPM for 5 sec and then 10 sec ramp, 3000 RPM for 40 sec), followed by a 1-minute soft-baking step at 115 °C. Electrode designs were then patterned into the photoresist using a $\mu PG101$ laser writer, (Heidelberg Instruments, Heidelberg, Germany) (Settings: 50 mW power, 60 % efficiency, 1 $\times$ 2 energy mode, –27 focus, 20 mm focal length writerhead.)

Once the drawing was complete, the exposed areas were removed

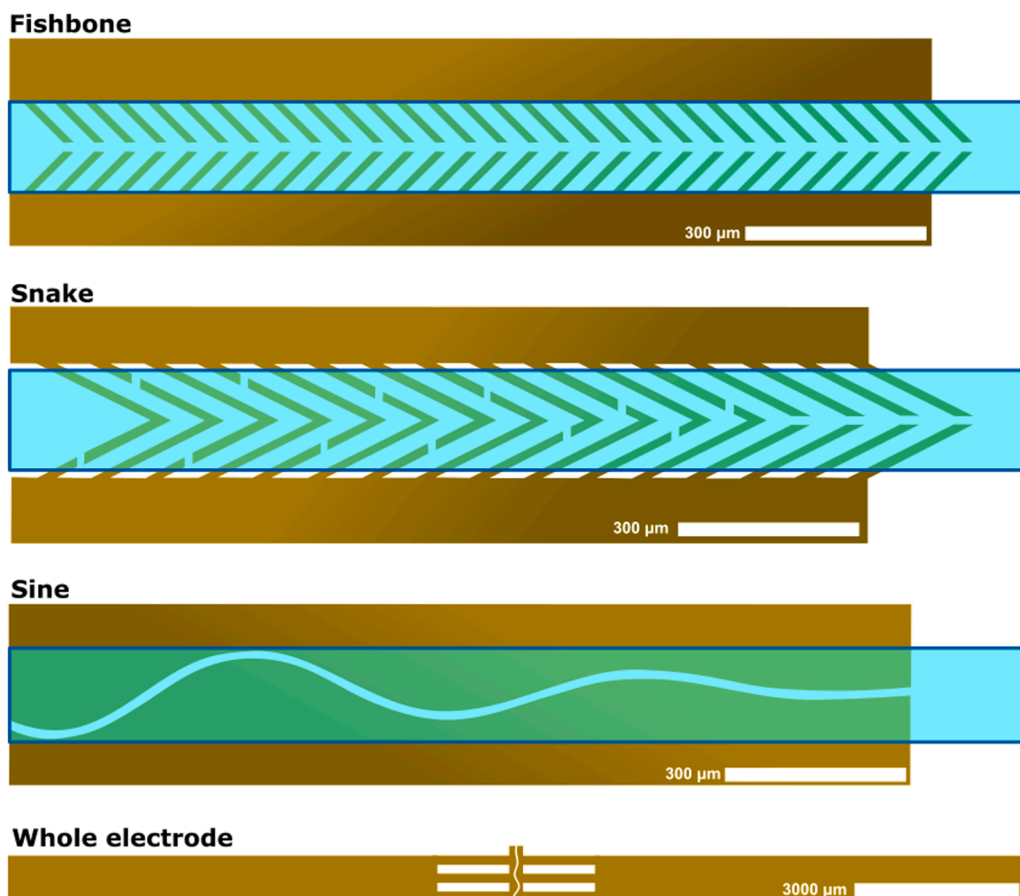


Fig. 2. The three electrode designs, the “fishbone,” the “snake,” and the “sine”. In the middle of each electrode, the blue region shows the microfluidic channel position. Below the three electrodes, the full design is shown, with the side-branches for the electric inlets.

using an MF-319 developer, leaving only the bare glass surface where the photoresist had been illuminated. Then the entire layer was subsequently exposed to UV light (Mercury i-line UV light, Oriel 97,435 UV lamp, $\lambda \approx 365$ nm), for a few seconds, to enable complete removal in the next development step.

Following the UV exposure, a few-nanometer thin gold layer was deposited by sputter-coating (Emitech K975X thermal evaporator). During the gold deposition, a current of 65 mA was applied for 13 s, with the substrate positioned approximately 40 mm from the gold source.

The gold-coated substrate was then immersed in an MF-319 developer overnight. This process lifted off the gold from regions still covered by the photoresist, leaving gold only on the previously exposed glass areas.

2.2. The microfluidic channel

The microfluidic channel was 100 μm wide and 10 μm high. Its layout featured a single inlet leading into a 1 mm long channel segment, where the particle-collecting electrode was positioned. After the electrode, the channel split into three branches corresponding to three outlets. The middle branch served as the particle-collection outlet, and the side channels were for the waste. The design was intended to be multifunctional. When only one particle type is present, then the electrode can concentrate the particles into the middle channel. In case of multiple particle sizes, the electrodes can focus one type of particle into the middle outlet, and the others into the side ones.

The mold for the channel was fabricated using the SU-8 2007 photoresist. The ca. 10–15-μm thin layer was made by spincoating a cover glass, using the following sequence: a 5-second ramp to 1000 RPM

for 5 s, followed by a 10-second ramp to 3000 RPM for 40 s. This is followed by a 3-minute 95°C heating hotplate. The channel pattern was drawn by the laser writer (settings: 50 mW power, 90 % efficiency, 1 × 2 energy mode, –15 focus, 4-mm focal length writerhead.) After a 4-minute bake at 95°C on a hotplate the plate was put into an SU-8 developer solution (mr-Dev 600, micro resist technology) for 3 min. Then the plate was washed with propanol that removed the unexposed area, and only the channel pattern for channel mold remained. The surface of the mold was treated with silane (TRIDECAFLUORO-1,1,2,2-TETRAHYDROOCTYL)TRICHLOROSILANE, Gelest) to prevent chemical bonding between the mold and the PDMS.

The microfluidic channel, created by PDMS (SYLGARD 184, Dow Chemical Company), made by using a mixture of PDMS pre-polymer and a curing agent in a 10:1 ratio. After mixing, it was vacuum-treated, which removed the bubbles from the mixture. It was then poured into the mold and cured by heating at 90 °C for approximately one hour, until solidified. Finally, the PDMS was carefully peeled off from the mold. For the inlet and the outlets, 1.2-mm holes were punched into the PDMS. Placing the PDMS and the electrode plate into a plasma cleaner (Plasma Cleaner / Sterilizer, Harrick Plasma) for one minute activates the surface of the PDMS, enabling it to form strong and watertight bonds with the glass. Applying oxygen plasma treating, the surface of the glass and the PDMS were oxidized, which allowed them to covalently bond to each other by forming O-Si-O bonds [21].

Finally, electrical connections were established by attaching wires to the electrode outlets using a CircuitWorks CW2400 conductive epoxy.

2.3. The electrode designs

In microchannels, the small cross-sectional dimensions (equivalent

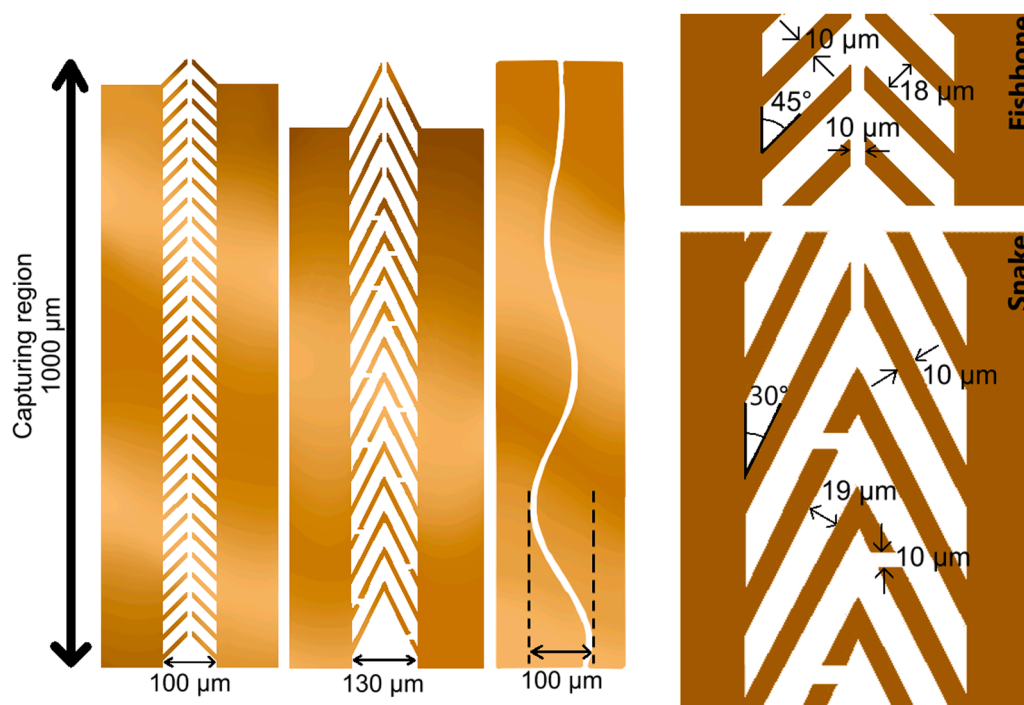


Fig. 3. The detailed figure of the electrode designs. The capturing region of all the electrodes are 1000- μm long and 100- μm wide. The electrodes of the “fishbone” design are 10 μm apart in the opposite sites, 18 μm apart on the same side, and tilted by 45° angle relative to the channel sidewall. The electrodes of the “snake” design are also 10 μm apart on the opposite sides, and 19 μm apart on the same side; they are tilted by 30° angle.

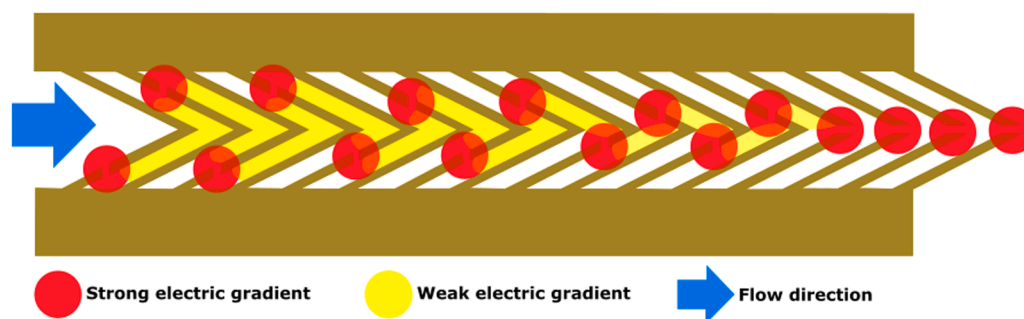


Fig. 4. Particle-trapping regions of the snake electrode. The red areas show the strongest electric gradient between the electrode ends. The yellow areas show the weaker gradient force between the oppositely charged edges. The blue arrow shows the direction of the fluid flow. The center of the first gap pair is 20 μm , while those of the subsequent pairs are 35 μm , 50 μm and 65 μm away from the side of the electrodes. The last section, where the gap is in the middle, has four repetitions.

hydrodynamic diameter) result in laminar flow, where streamlines run parallel to the channel walls. There are no vortices, and therefore there are no forces acting in lateral direction on any of the liquid elements, perpendicular to the streamlines. For microparticles suspended in a liquid, the inertial forces are much smaller than the drag force, so they move at the same speed as their liquid environment. Therefore, in laminar flow the dragged microparticles follow the streamlines. To change the lateral position of such a particle, an external force is needed, which in our device is provided by DEP. That force can be an attraction toward the highest gradient of the alternating electric field (pDEP), or can be the opposite case (nDEP) when the highest gradient repulses the particle. In our device we apply pDEP. Pointed electrode pairs with small curvature radius generate the strongest field gradients, whereas electrodes with larger curvature radius produce weaker gradients. For an electrode pair separated by a gap, the resulting gradient is perpendicular to the edge of the electrodes. In the case of curved edges, the effective gradient is defined by the sum of these local perpendicular components (Fig. 2).

For efficient particle collection, two conditions must be satisfied by

the pDEP forces acting on the particles. First, a force component must exist perpendicular to the channel walls (and therefore to the flow direction). Second, this lateral force component should always point toward the centerline of the channel at each position. Under these conditions, once a particle reaches the centerline, it remains stably confined there.

We designed three distinct electrode patterns: the “fishbone”, the “snake”, and the “sine” structures, shown in Fig. 3. The overall construction of the electrode structures were similar: they were 1-mm long, and consisted of alternately changing positive and negative electrode series placed on opposite sides of the channel. They were designed to concentrate the particles at the center of the channels. For electrical connection, a roughly 10-mm long gold side branch had been placed on both sides of the electrodes, where wires could be connected. The electrodes were made of a few-nm thick gold layer.

2.3.1. Fishbone

The first electrode design was the “fishbone”, named after its structure. On both sides of the channel, a series of electrodes was tilted at a

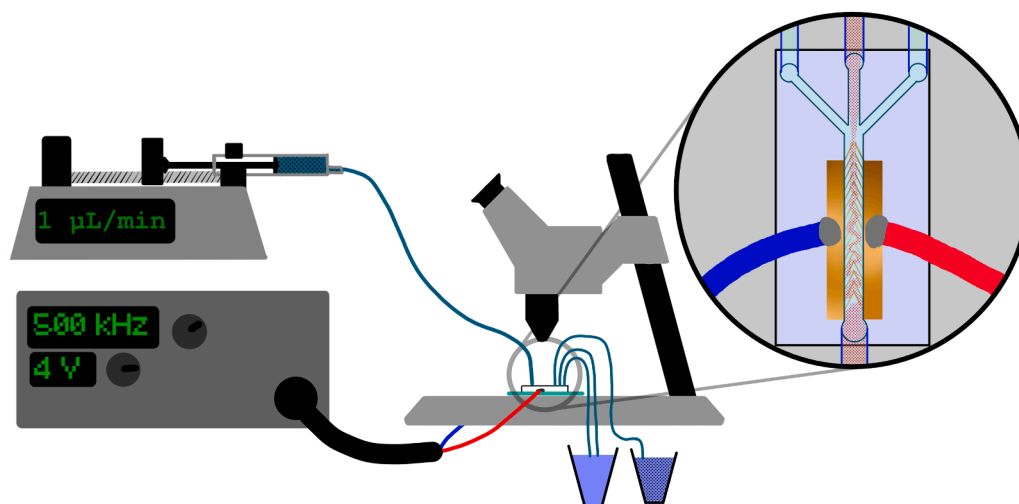


Fig. 5. The design concept of the particle-collecting system. Fluid flow is regulated by a syringe pump, while voltage is supplied by a function generator. The electrodes capture the microbeads, direct them into the middle collecting outlet, and the waste goes to the side outlets. The electrode system (the figure shows the snake design during operation) was under a microscope, so that it could be observed in action.

given angle related to the sidewall of the channel. This design was used in one of our group's previous projects [20]. The best angle for particle collection was identified by COMSOL simulations. To maximize the DEP force's lateral effect, 45°-angled electrodes were selected for the final design.

The highest electric gradient is at the electrodes' ends. Also the edges of the opposing electrodes generate additional gradients that guide particles along these straight edges. Since the adjacent electrode edges are tilted toward the channel centerline, they direct the particles toward it, where the highest gradient enables their laterally constrained tangential flow.

The detailed final design is shown in Fig. 3. The angle of inclination of the electrodes is 45°, the distance between the electrodes on the same side is 18 µm, and the distance between the opposite-side electrodes is 10 µm. The electrodes are 1000-µm long, and 100-µm wide, which is the full width of the channel.

2.3.2. Snake

The "snake" design is the improved version of the "fishbone" configuration. The main objective of this new design is to enhance the lateral effect of the electrodes. The detailed design shown in Fig. 3.

This electrode system consists of parallel, equidistant, „V" shaped lines with one gap on each. That gap separates the two sides and forms counter-electrode pairs. The highest gradient appears at the narrowest gaps. These regions can be considered as trapping zones. As it is shown in Fig. 4, these trapping places are organized in lines that are symmetric to the centerline of the channel. Two relevant forces act on the particles there, the pDEP force and the drag force of the flow. Only the first force has lateral components, while the other one is stronger, but acts only in the tangential (flow) direction. In other words, the laminar flow is parallel to the channel, only the DEP force can distract the particles' trajectories. The gaps between the electrodes are placed to direct the particles into the middle of the channel.

We designed an electrode geometry according to this concept. The change between the subsequent electrode pairs is the location of the gap between them. These gaps alternate between the left and right sides. They start near the wall of the channel and extend toward its center. For improved efficiency, each left-right gap pair is repeated twice. Each trapping zone of the consecutive electrode pairs is one step closer to the centerline, slightly overlapping with the previous zone. This design is also able to weakly trap particles between the edges of the oppositely charged electrodes. The center of the first gap pair from the side of electrodes is 20-µm, while the subsequent pairs are 35 µm, 50 µm and

65 µm away from the side of the electrodes. The last section, where the gap is in the middle, has four repetitions.

The distance between the two outer sides of the electrode series is 130 µm, allowing for easier placement of a 100 µm-wide microfluidic channel. The full length of the snake electrode is 1000 µm. Each electrode is angled at 30° related to the sidewall of the channel, with a spacing of 19 µm between electrodes on the same side, and 10 µm between the electrodes on opposite sides.

While the fishbone design determined 45° as the optimal angle for lateral particle collection, this snake design uses 30° angles. This is because the placement of the gaps already provides a strong lateral DEP effect, and this angle makes it easier to remove the unnecessary gold from the area between the electrodes, during the preparation step.

The design is named "snake" because the particles exhibit a snake-like trajectory during operation.

2.3.3. Sine

The last electrode design is the "sine". This design differs from the other two, as it has a big electrode pair, not a series of small, parallel stripes. The gap between the sides is constant, 10 µm, and it follows a sine curve with decreasing amplitude towards the end of the channel. The curve is designed to traverse the entire width of the channel. The length of the electrode area is 1000 µm, and the maximum amplitude of the curve is 100 µm. The design is shown in Fig. 3. This electrode system can be considered as the continuous version of the snake design, because here the gap (i.e. the capturing region) approaches the centerline gradually (and not in discrete steps).

The highest gradient of the electric field in this electrode system is perpendicular to the edge of the electrodes, so the strongest component of the pDEP force is always perpendicular locally to the line (edge) of the gap. As a result of that, the particles dragged by the flow tend to follow this damped sine wave that leads them to the middle. Hence, the particles follow a snake-like crawling line tending to the middle, such as in the snake electrode system, which inspired the shape of the electrodes. A practical advantage of the sine electrode design that it does not require such a high spatial resolution as needed for producing the comb-like thin parallel lines of the snake electrode.

2.4. Experimental setup

Fig. 5 illustrates the experimental setup for the particle collection system. The fluid flow was provided by a syringe pump (Legato 100, KD Scientific) through a PTFE tube of 0.3 mm inner diameter, connected to

the microfluidic channel. After the channel was filled with the solution containing the microbeads, the voltage was applied using an electric-signal generator (Function Generator 8020, Tabor Electronics Ltd.). The electrodes captured the particles in the middle of the channel and directed them toward the channel outlet. The final channel design featured one inlet and three outlets: two side outlets for waste and one central outlet for particle collection. For the tests we only used channels with one or two outlets for the sake of simplicity. The chip was placed under a microscope to enable real-time observation during operation.

3. Results and discussion

3.1. Simulations

Two types of particles were used: 1 μm polystyrene microbeads (Polybead Microspheres 1.00 μm , Polysciences) and 0.1 μm fluorescently stained latex microbeads (Red Fluorescent Carboxylated PS Latex, 0.10 μm 2.5 %, Magsphere). The particles' polydispersity index was below 0.1, as measured by a Zetasizer (Malvern Panalytical), meaning that their size distribution was very narrow, i.e. their aggregation was minimal.

For the DEP force simulations we used COMSOL Multiphysics simulation software (Supplementary item 1). The time-averaged force was calculated with the following formula, assuming that the particles can be considered spherical:

$$F_{DEP} = 2\pi r^3 \epsilon_0 \text{Re}(\epsilon_m^*) \text{Re}(CM) |\nabla E|^2, \quad (2)$$

where CM is the Clausius-Mossotti factor ($CM = 0.28177 - 0.1567i$):

$$CM = \frac{\epsilon_p^* - \epsilon_m^*}{\epsilon_p^* + 2\epsilon_m^*}, \quad (3)$$

r is radius of the particle,

- ϵ_0 is the vacuum permittivity,
- ϵ_m^* is the medium complex permittivity,
- ϵ_p^* is the particle complex permittivity,
- E is the electric field.

For the calculations, the medium bulk conductivity and the zeta potential of the particles were measured by a zeta-potential meter (Zetasizer, Malvern Panalytical).

- 1 μm particle zeta potential: -0.04 V
- 0.1 μm particle zeta potential: -0.03 V
- Water bulk conductivity for the 1 μm particles: 0.005 S/m
- Water bulk conductivity for the 0.1 μm particles: 0.0065 S/m

For the complex permittivities, the following formula were used:

$$\epsilon^* = \epsilon - \frac{i\sigma}{\omega\epsilon_0}, \quad (4)$$

where

- ϵ is the dielectric constant (for water $\epsilon_m = 80$, for the particles $\epsilon_p = 2.6$),
- i is the imaginary unit,
- σ is the bulk conductivity of the medium and effective conductivity of the particle,
- ω is field frequency ($\omega = 3.14$ MHz).

The polystyrene particles' conductivity is approximated by its effective surface conductivity:

$$\sigma_{p,eff} = \sigma_p + \frac{2\sigma_s}{r}, \quad (5)$$

where σ_s is the surface conductivity of the particle:

$$\sigma_s = \frac{4F^2 I z^2 D (1 + 3m/z^2) \lambda_D}{RT} \left(\cosh \frac{zF\zeta}{2RT} - 1 \right), \quad (6)$$

- σ_p is the particle conductivity,
- r is particle radius,
- F is the Faraday constant,
- I is the ionic strength,
- z is the ion valency ($z = 1$),
- D is the diffusion coefficient,
- R is the gas constant,
- T is the absolute temperature ($T = 298$ K),
- λ_D is the Debye length,
- ζ is the zeta potential,
- m is the contribution of electro-osmosis:

$$m = \frac{2\epsilon_0 \epsilon_m R^2 T^2}{3\eta F^2 D}, \quad (7)$$

where η is the dynamic viscosity of water.

The diffusion coefficient is:

$$D = D_{OH^-} + D_{H^+}, \quad (8)$$

where

- D_{OH^-} and D_{H^+} are the diffusion coefficients of the OH^- and H^+ ions.
- The Debye length is:

$$\lambda_D = \sqrt{\frac{\epsilon_0 \epsilon_m k_B T}{2e^2 N_A I}}, \quad (9)$$

where

- e is the elementary charge,
- N_A is the Avogadro number.

Ionic strength calculated by the following formula:

$$I = \sum_i z_i c_i, \quad (10)$$

where

- z is the valency of ion i ,
- c is the concentration of ion i .

We used water as medium, so there were only OH^- and H^+ ions in the solution.

Therefore $z_{OH} = z_H = 1$ and $c_{OH} = c_H = c$. With these simplification the formula of the ion strength is the following:

$$I = c_{OH^-} + c_{H^+} = 2c \quad (11)$$

Finally, the concentration calculated from the bulk conductivity of the medium:

$$\sigma_m = \sum_i c_i z_i^2 e^2 \frac{D_i}{k_B T} \quad (12)$$

From these, c can be derived:

$$c = \frac{k_B T \sigma_m}{z^2 e^2 N_A (D_{OH^-} + D_{H^+})}, \quad (13)$$

In summary, the following values were used for the DEP force calculations for the 1 μm polystyrene and the 0.1 μm latex particles:

Although the two particle types exhibit similar bulk conductivity, their effective conductivities differ. Since the particles are insulators,

Table 1

The base values for the DEP force calculations.

	1 μm particle	0.1 μm particle
Complex permittivity of medium	80–179.751i	80–233.6763i
Bulk conductivity of medium	0.005 S/m	0.0065 S/m
Complex permittivity of particle	2.6–426.7842i	2.6–4058.0126i
Effective particle conductivity	0.0118 S/m	0.1128 S/m
Frequency	500 kHz	500 kHz

their conductivity is negligible, making surface conductivity the dominant contribution. In this case, the effective surface conductivity is inversely proportional to the particle radius, hence the one order of magnitude difference in their effective conductivity, as well.

Note that in the zeta potential measurements, the standard

deviations are 0.00339 V and 0.00367 V for the 1 μm and the 0.1 μm particles, respectively. Hence, the respective calculation uncertainties are 18 % and 2 % for the 1 μm and 0.1 μm particles. The relative errors from the uncertainties of the bulk conductivity measurements are 12 % for the 1 μm particles (standard deviation = 0.0006 S/m) and 3 % for the 0.1 μm particles (standard deviation = 0.001 S/m). (See Supplementary items 2–3.)

3.1.1. Enhancement of the DEP force range

We conducted five simulation series to evaluate how changes in certain parameters affect the range of the DEP force. In the following simulation the scale of the DEP force was maximized at 10×10^{-13} N for better visibility.

These tests were performed using a simplified electrode design

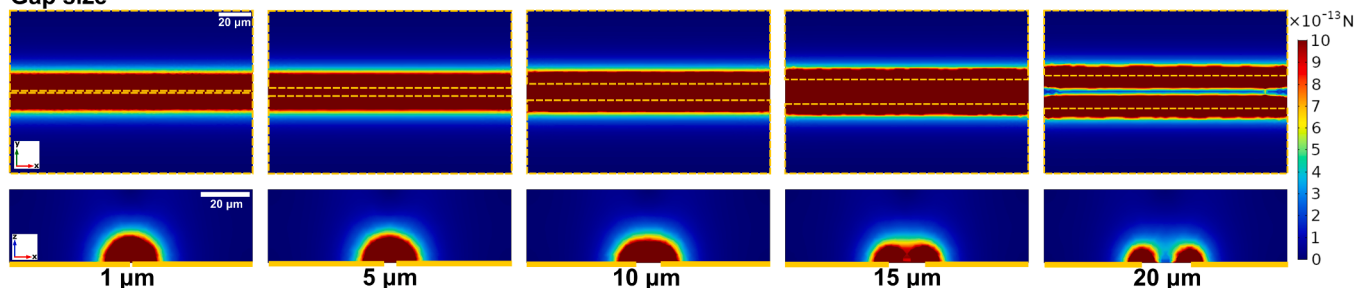
Gap size

Fig. 6. DEP force characterization for the simplified electrode design: this simulation series illustrates the dependence of the DEP force on the electrode gap size (1, 5, 10, 15 and 20 μm). Each column includes two views: a top view (upper row) and a cross-sectional view perpendicular to the gap (lower row). The gold dashed line in the top view, and the simple line in the cross-sectional view show the positions of the electrodes. The scale bars represent 20 μm , while the magnitude of the force was maximized at 10×10^{-13} N for better clarity.

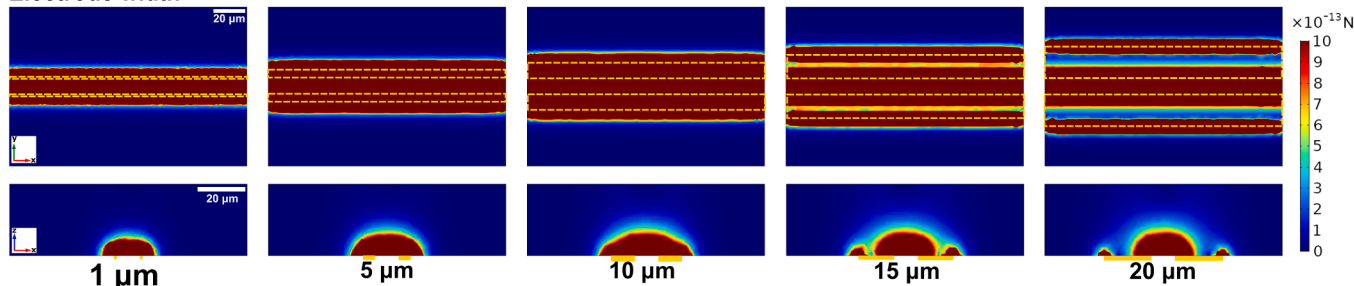
Electrode width

Fig. 7. DEP force characterization for the simplified electrode design: this simulation series illustrates the dependence of the DEP force on the electrodes' width (1, 5, 10, 15 and 20 μm), while keeping the gap size constant (10 μm). Each column includes two views: a top view (upper row) and a cross-sectional view perpendicular to the gap (lower row). The gold dashed line in the top view, and the simple line in the cross-sectional view show the positions of the electrodes. The scale bars are 20 μm , and the magnitude of the force was maximized at 10×10^{-13} N for better clarity.

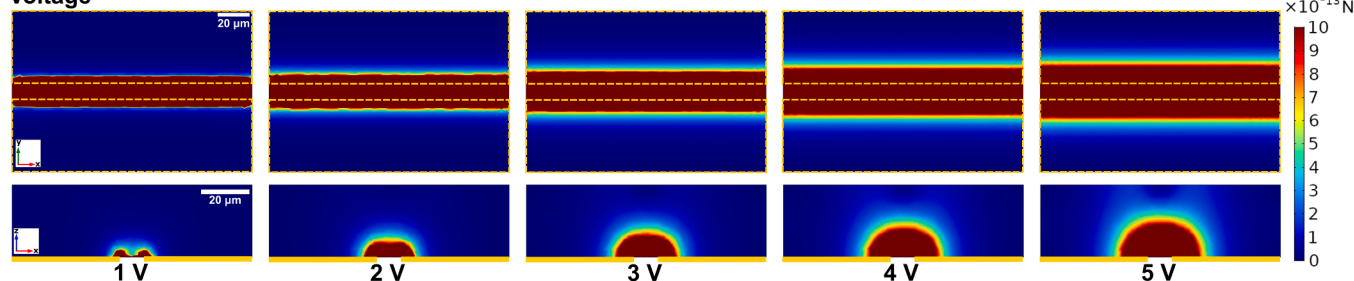
Voltage

Fig. 8. DEP force characterization for the simplified electrode design: this simulation series illustrates the dependence of the DEP force on the applied voltage (1, 2, 3, 4 and 5 V). Each column includes two views: a top view (upper row) and a cross-sectional view perpendicular to the gap (gap size 10 μm), (lower row). The gold dashed line in the top view and the simple line in the cross-sectional view show the positions of the electrodes. The scale bars stand for 20 μm , and the magnitude of the force was maximized at 10×10^{-13} N for better clarity.

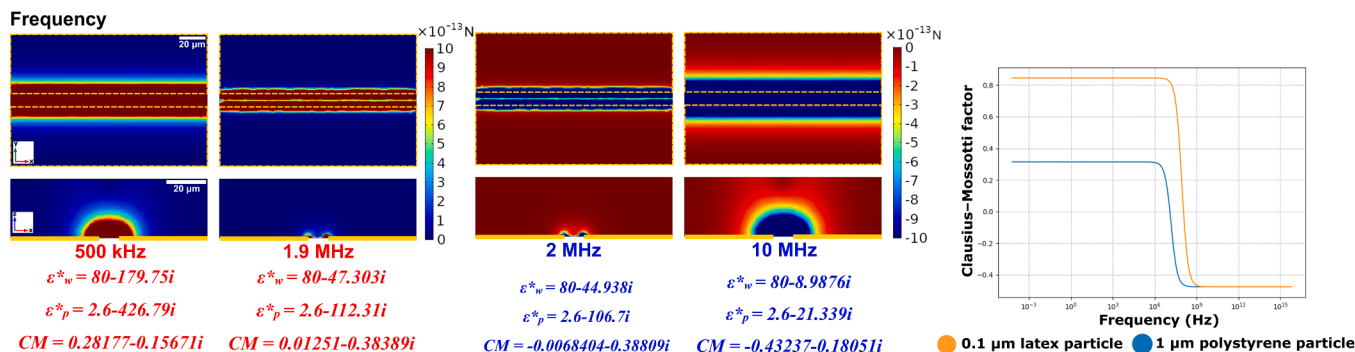


Fig. 9. DEP force characterization for the simplified electrode design: this simulation series illustrates the dependence of the DEP force on the frequency (0.5, 1.9, 2 and 10 MHz). Each column includes two views: a top view (upper row) and a cross-sectional view perpendicular to the gap (gap size 10 μm), (lower row). The gold dashed line in the top view and the simple line in the cross-sectional view show the positions of the electrodes. The scale bars are of 20 μm, and the magnitude of the force was maximized at 10×10^{-13} N for better clarity. Below the maps, the frequency-dependent permittivity values and the Clausius-Mossotti factors are shown for the polystyrene beads. The red numbers indicate positive, while the blue ones negative DEP. The last column shows how the Clausius-Mossotti factor depends on the frequency for the 1 μm polystyrene and 0.1 μm latex particles, respectively.

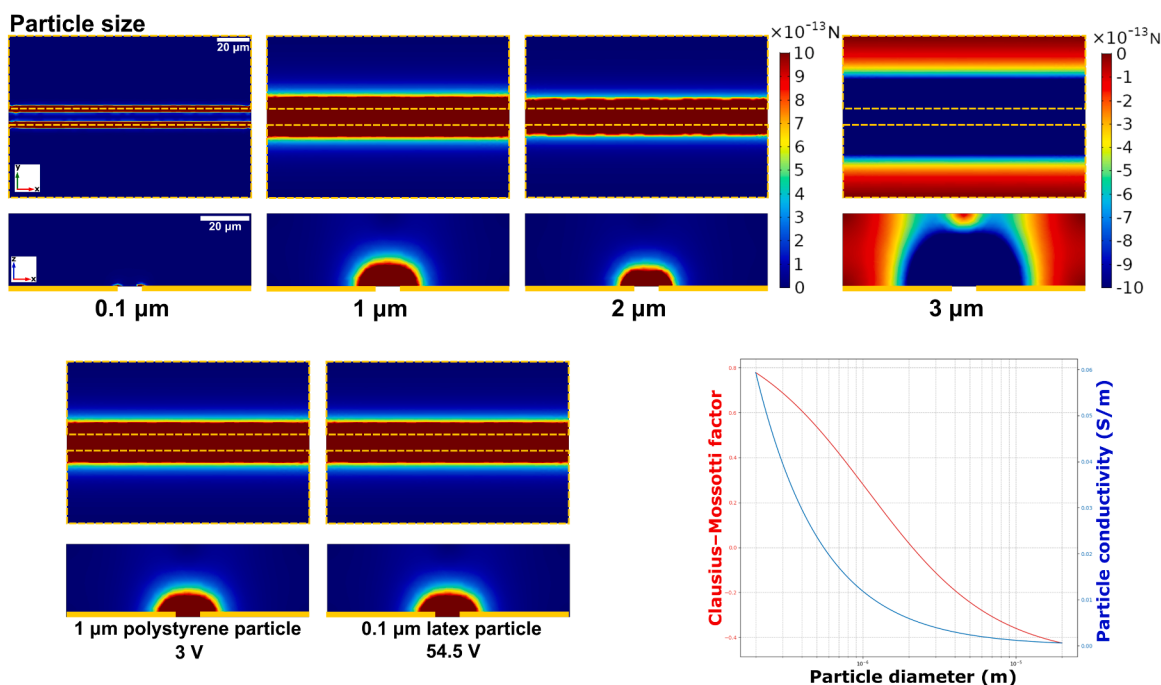


Fig. 10. Characterization of the DEP force for the simplified electrode design. This simulation series illustrates the dependence of the DEP force on the particle size (0.1, 1, 2 and 3 μm). The upper rows show the top views, while the lower rows are the cross-sectional views perpendicular to the gap. The lower left inserts show a comparison for the voltage required to achieve the same DEP-force range for the 1 μm polystyrene particles and the 0.1 μm latex particles. The gold dashed line in the top view and the simple line in the cross-sectional view show the position of the electrode. The scale bar is 20 μm, and the magnitude of the force was maximized at 10×10^{-13} N for better clarity. Next to the simulations the graph shows how the effective conductivity and hence the Clausius-Mossotti factor depends on its size.

consisting of a single electrode pair. In the base configuration, the electrodes spanned the entire channel, separated by a 10 μm longitudinal gap parallel to the microfluidic channel. The figures show the simulations for the 1 μm polystyrene particles. We used the parameters of Table 1, and changed only one parameter per simulation series (gap size, electrode width, voltage, frequency, particle size). The voltage for the particle collection was set to 3 V, unless specified otherwise.

The following figures (Figs. 6–10) show results of the simulated systems in top view (upper rows) in the plane of the electrodes, and in cross-sectional view (lower rows), perpendicular to the gap.

In the first series of simulations, we varied the gap between the electrodes, using distances of 1, 5, 10, 15 and 20 μm (Fig. 6 upper row). The DEP force field was formed along the sides of the electrodes. For small gaps, the opposing electrode edges were close enough for their

DEP fields to merge. The merging force fields can enhance each other's lateral range, which is why the effective DEP range was slightly larger for the 1 μm particles than for the 5 μm particles. Above that, as gap size increased, the lateral range of the force increased as well, until a split of the field occurred. At 20 μm gap size, the electrodes are too far apart to merge their DEP force fields, and without this, the range of the DEP force was reduced, as seen in the corresponding image. This separation results in the formation of two distinct particle trapping zones at 20 μm gap size. The range of the DEP force above the electrodes also changed by gap size (Fig. 6 lower row): where overlapping, the fields reinforced each other vertically, as well. In summary, a careful selection of the electrode gap size can slightly enhance the effective range of the DEP force.

In the second simulation series, the lateral width of the electrodes

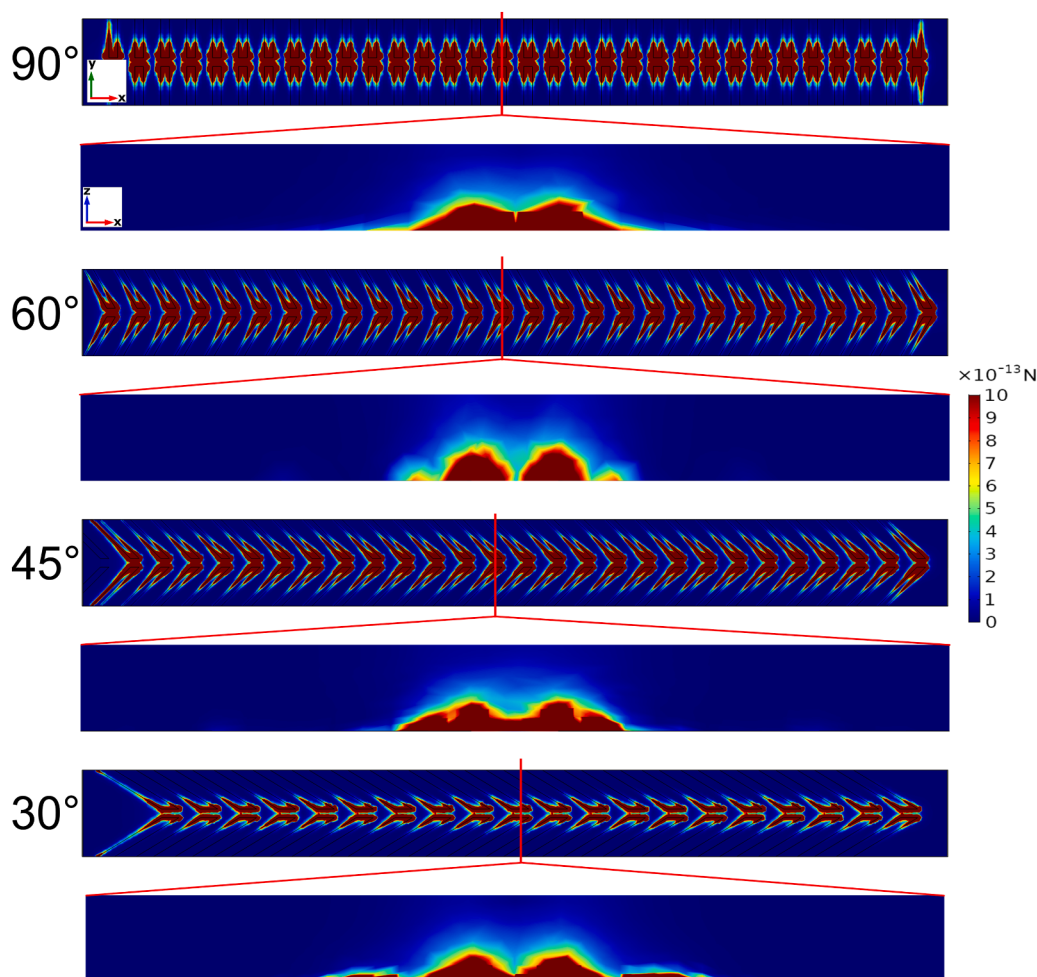


Fig. 11. DEP force simulations for fishbone electrode designs. For the simulation $1\ \mu\text{m}$ polystyrene particles and $1\ \text{V}$ was used. Each design has differently tilted electrodes (90° , 60° , 45° and 30° relative to the channel sidewall). The designs have two views: a top view from the plane of the electrode, and a cross-sectional view from the red line in the image. The same scale, maximized at $10 \times 10^{-13}\ \text{N}$ for better clarity, stands for each simulation.

was varied (Fig. 7) which is relevant for the fishbone and the snake designs. In these configurations, the electrodes had widths of 1 , 5 , 10 , 15 , and $20\ \mu\text{m}$, separated by a $10\text{-}\mu\text{m}$ gap, shown by the gold contour lines. Narrowing the electrodes reduces the vertical DEP range, contrary to the gap size which can enhance this range, if it is small enough. DEP force was generated along both sides of each electrode. When the two sides were close enough, their DEP fields overlapped, extending the lateral range approximately to the width of the electrode, as seen for electrodes 1 , 5 , and $10\ \mu\text{m}$ wide. However, when the sides are too far from each other, then two new, smaller trapping zones are formed. In summary, narrow electrodes reduce the vertical DEP range, and the lateral range increases with the electrode width, while excessively wide electrodes generate distinct, unfavorable trapping zones.

In the third simulation series the voltage was changed (Fig. 8). Simulations were performed at 1 , 2 , 3 , 4 and $5\ \text{V}$. The voltage and the DEP range (both laterally and vertically) increased together. Note, however, that in practice the larger voltages might be harmful for the biological particles and the electrodes.

Another parameter which influences the DEP range is the frequency, via the frequency dependence of the complex permittivities of the particle and the medium. The Clausius-Mossotti factor is determined by the permittivities and the polarizabilities of the particles. If CM is positive, the DEP is also positive (pDEP), while negative CM implies a negative DEP (nDEP). Our electrodes were optimized to pDEP, where the force field attracts the particles. (In the case of nDEP, the DEP field repels the particles.) The frequency-dependent simulations are presented in Fig. 9.

The frequency-dependence of CM can be divided into three sections: an approximately constant positive section, a short steep section where the sign changes, and finally an approximately constant negative section [22]. In Fig. 9, the map taken at $500\ \text{kHz}$ demonstrates the positive section, the $1.9\ \text{MHz}$ and $2\ \text{MHz}$ maps show the positive and negative sides of the sign-change section, respectively, while the $10\ \text{MHz}$ map shows the negative section. The rightmost column of Fig. 9 shows the frequency dependence of the Clausius-Mossotti factor for both the $1\ \mu\text{m}$ polystyrene and $0.1\ \mu\text{m}$ latex particles. CM of the smaller particles spans larger positive values, and the two particle types cross zero at different frequencies: the smaller at $1.950847\ \text{MHz}$, while the larger at $18.528813\ \text{MHz}$. This could be utilized in an effective separation of these particles by DEP. (For instance, at $10\ \text{MHz}$, the $1\ \mu\text{m}$ particles experience negative DEP, whereas the $0.1\ \mu\text{m}$ particles experience positive DEP.)

Finally, the particle-size dependence of DEP was simulated (Fig. 10). Although the formula of the DEP force is proportional to the cube of the particle diameter, it does not increase with this factor, because the surface conductivity of the particle is inversely proportional to its diameter. As a result, the Clausius-Mossotti factor also exhibits a more moderate dependence on particle size. Simulations were performed for 0.1 , 1 , 2 and $3\ \mu\text{m}$ polystyrene particles. For particles larger than $2\ \mu\text{m}$, the DEP response becomes negative due to the reduction in effective particle conductivity. The size-dependent behavior of both conductivity and the Clausius-Mossotti factor is illustrated in Fig. 10.

In addition, the bottom rows of Fig. 10 compare $1\ \mu\text{m}$ polystyrene

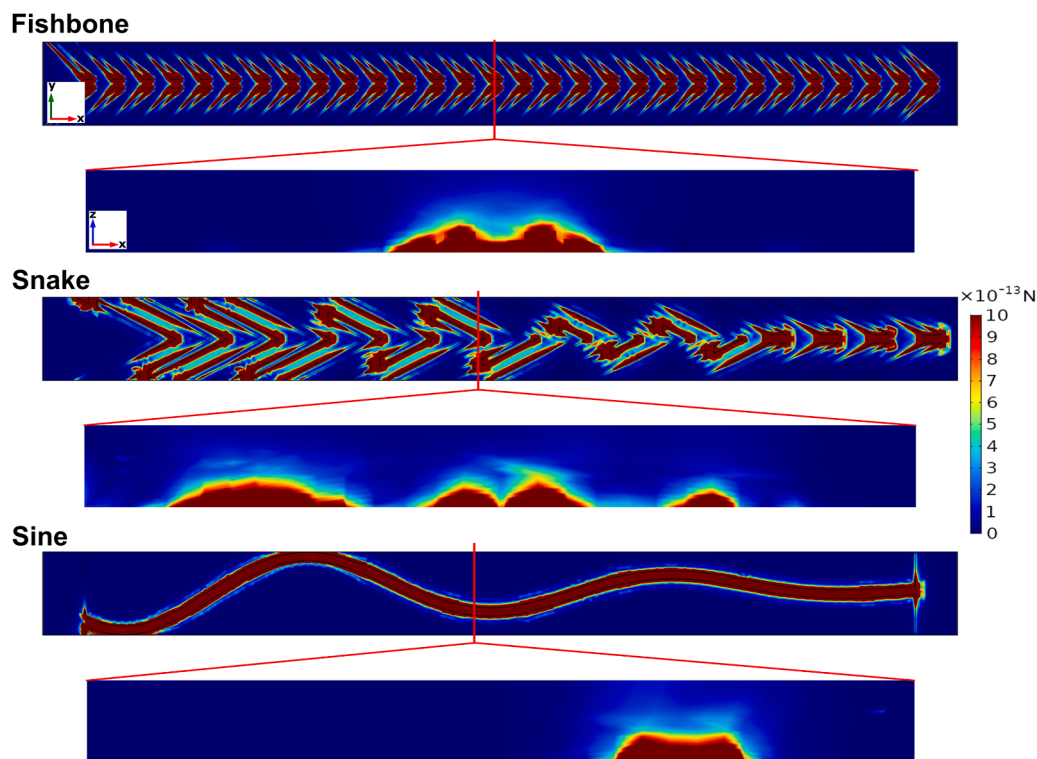


Fig. 12. DEP force simulations for the three final electrode designs. For the simulation 1 μm polystyrene particles and 1 V was used. Each design has two views: one top view, from the plane of the electrode and a cross-sectional view from the cross-section at the red line in the image. The same scale, maximized at 10×10^{-13} N for better clarity, stands for each simulation.

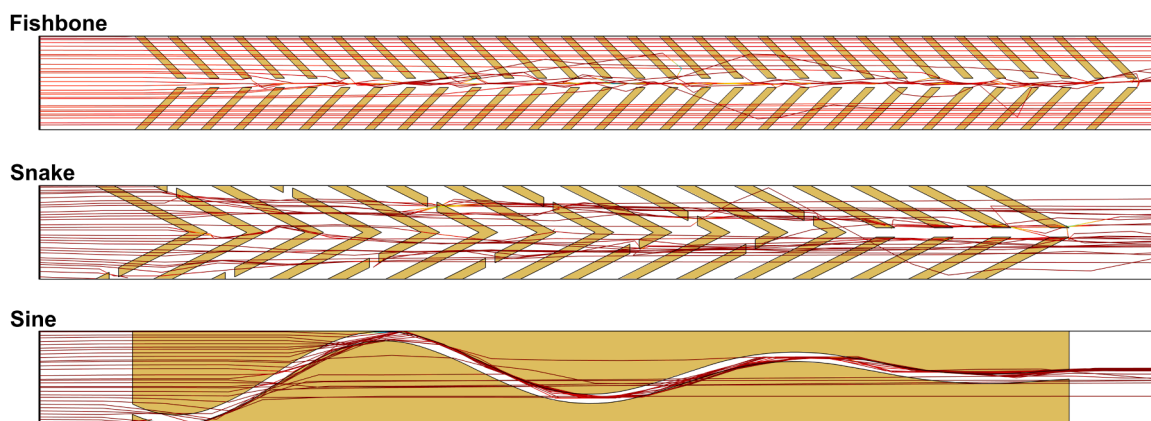


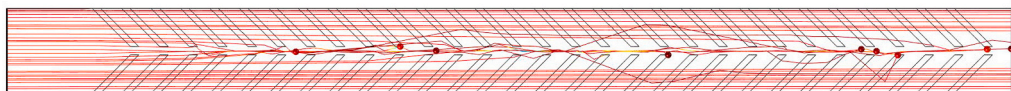
Fig. 13. Particle-tracing simulation in COMSOL. The red lines show the traces of the particles during the simulation. In the right end of the channels the blue region shows the collection area. To maximize the collecting efficiency, different voltages were applied for each design (fishbone 3 V, snake 4.5 V, sine 4 V). 30 particle trajectories were simulated for each design. The particles started at the middle plane of the inlet, at 7–8 μm height. The particle-tracing simulation videos are shown in supporting videos ([Supplementary items 4–6](#)).

particles with 0.1 μm latex particles, highlighting the difference in the applied voltage required to achieve the same DEP force range. The results show that the smaller particles require approximately 18 times higher voltage to produce an equivalent DEP force. Under stationary conditions, however, the DEP force keeps balance with the Stokes drag force ($F_{\text{Stokes}} = 6\pi\eta r v_n$, where η is the viscosity and v_n is the normal component of particle speed, pointing toward the electrode), so the ultimate DEP effect in terms of speed is only 1.8 times higher for the 1 μm particles than for the 0.1 μm ones, provided that the same voltage is applied. Indeed, experimental results are in line with this conclusion, inasmuch as effective collection of the 0.1 μm particles require only about twice as high voltage as used for the 1 μm ones (10–15 V versus 4–7 V, respectively).

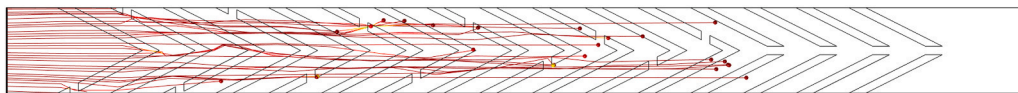
3.1.2. Fishbone design, electrodes' angle

In this simulation series, the optimal electrode angle for the fishbone configuration was identified by studying the DEP force for 90°, 60°, 45° and 30° electrode angles relative to the channel sidewall. This is shown in Fig. 11. Each design has two views: a top view in the plane of the electrode, and a front view in the cross-section at the red line in the image. In this simulation series, 1 μm polystyrene particles and 1 V was used to make the range of the DEP force more perceptible. The other simulation parameters remained unchanged.

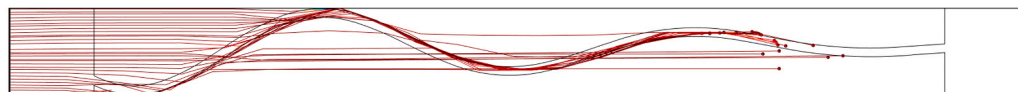
The 60°- and the 45°-tilted electrodes had a few- μm larger lateral DEP-force range, while the vertical range was about the same. Therefore, to optimize the lateral effect, the 45°-angled electrodes were selected for the experiments.



Video 1. Particle tracing simulations for the fishbone electrode. A video clip is available online. Supplementary material related to this article can be found online at [doi:10.1016/j.colsurfa.2026.140045](https://doi.org/10.1016/j.colsurfa.2026.140045).



Video 2. Particle tracing simulations for the snake electrode. A video clip is available online. Supplementary material related to this article can be found online at [doi:10.1016/j.colsurfa.2026.140045](https://doi.org/10.1016/j.colsurfa.2026.140045).



Video 3. Particle tracing simulations for the sine electrode. A video clip is available online. Supplementary material related to this article can be found online at [doi:10.1016/j.colsurfa.2026.140045](https://doi.org/10.1016/j.colsurfa.2026.140045).

3.1.3. Fishbone-snake-sine comparison

A major limitation of the fishbone electrode design is its restricted lateral effective range. The snake and sine designs were created to compensate for this. In the initial stage of the snake electrode, the DEP force extends to the full width of the electrode system, and then the area of the DEP gradually decreases, while directing the particles towards the centerline, until it only covers only central region of the channel. The sine design has a similar concept: at start, the wave traverses the entire width of the channel, before gradually “damped” to the middle of the channel. Both the snake and the sine designs are scalable for any channel width: the snake can be complemented by other gap pairs, while the wave in the sine design can also be extended at will, so they can be spread out for the wider channel. On the contrary, fishbone electrodes are only working well in narrow channels. The comparison for the

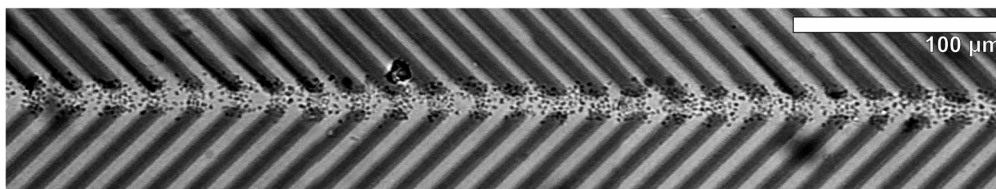
electrode designs is shown in Fig. 12. Similarly to the previous simulation series, 1 V voltage was used to make the range of the DEP force more perceptible.

3.1.4. Particle tracing simulation

After the DEP force range simulations, the next step was to calculate the particle-collecting efficiency of the electrodes. To get these efficiencies, the dielectrophoretic particle-collection process was simulated. For the simulations, the 0.1 $\mu\text{L}/\text{min}$ fluid rate and the Table 1 parameters were used.

To maximize the collecting efficiency, different voltages were applied for each design. The values were set to minimize permanent trapping, while collecting as many particles as possible. For the fishbone design we used 3 V, for the snake 4.5 V, and 4 V for the sine, the other

Fishbone



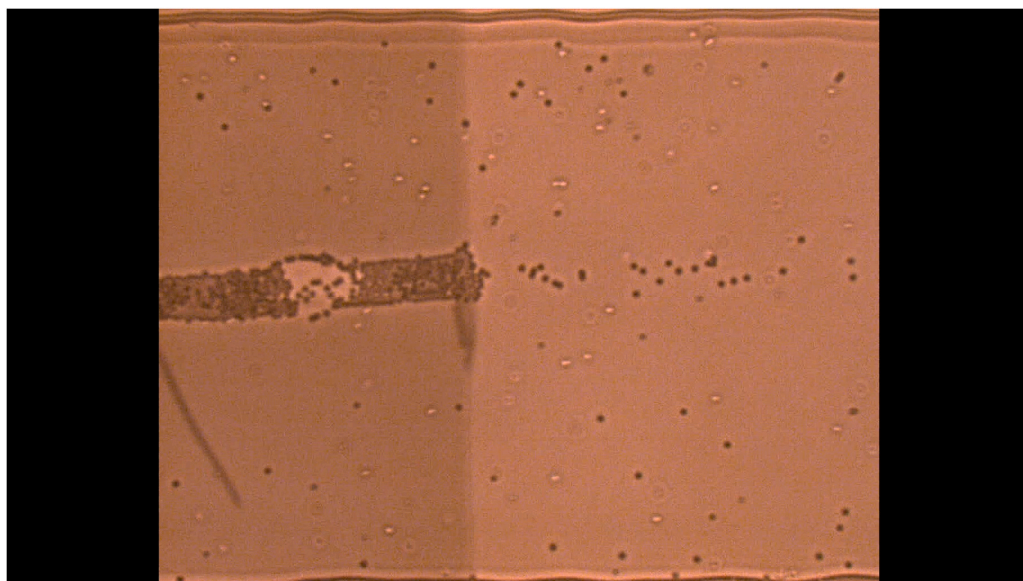
Snake



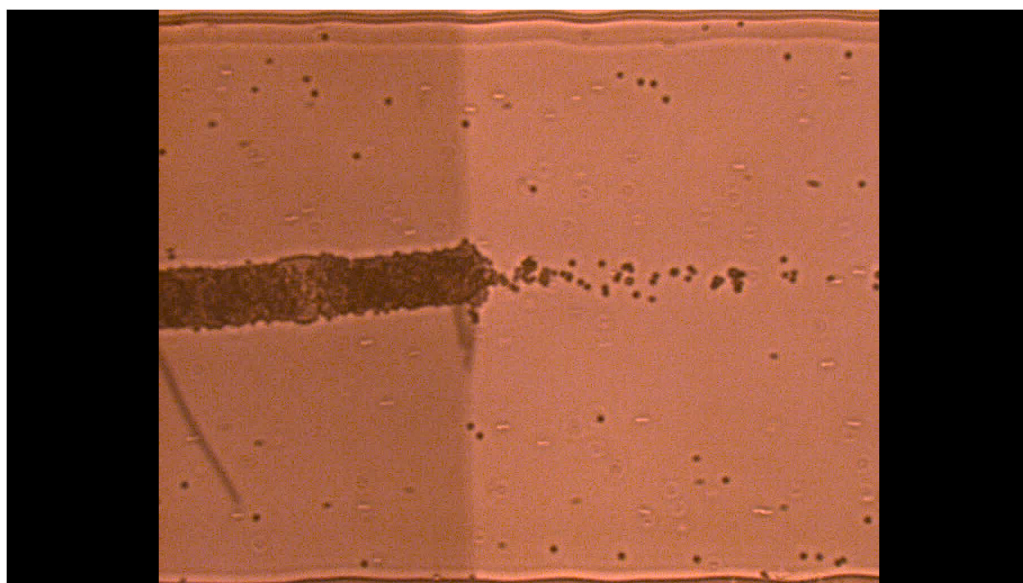
Sine



Fig. 14. Fishbone, snake and sine electrodes during operation. The diameter of the collected particles is 1 μm .



Video 4. Particle collecting experiments for 1 μm particles with the sine electrode at a flow rate of 0.005 $\mu\text{L}/\text{min}$. A video clip is available online. Supplementary material related to this article can be found online at [doi:10.1016/j.colsurfa.2026.140045](https://doi.org/10.1016/j.colsurfa.2026.140045).



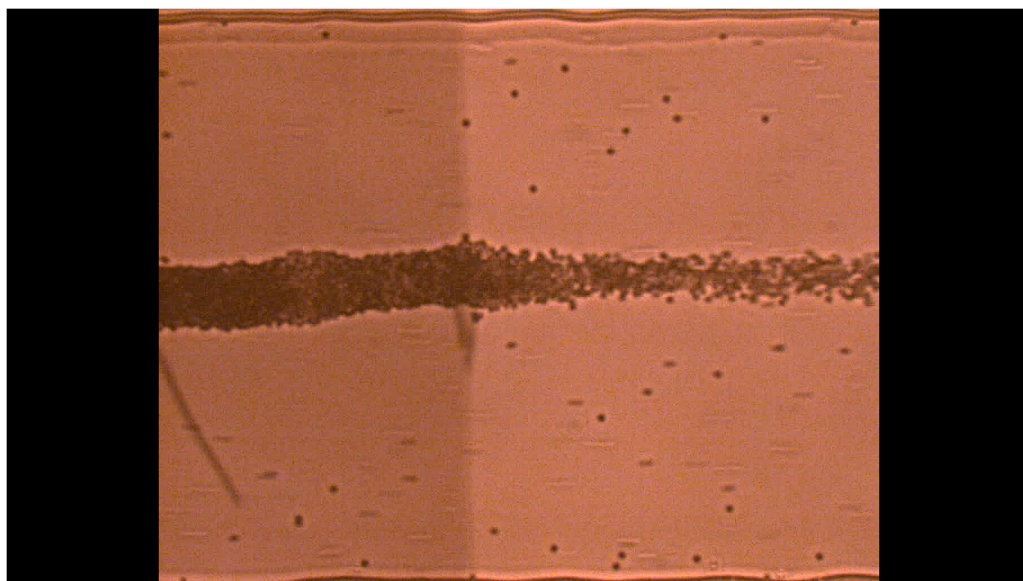
Video 5. Particle collecting experiments for 1 μm particles with the sine electrode at a flow rate of 0.015 $\mu\text{L}/\text{min}$. A video clip is available online. Supplementary material related to this article can be found online at [doi:10.1016/j.colsurfa.2026.140045](https://doi.org/10.1016/j.colsurfa.2026.140045).

parameters remained the same. We simulated 30 particles in each case, and ran each simulation 3 times, so overall we had 90 simulated particles for each electrode. The particles started at the middle plane of the inlet, 7–8 μm high. Particle-tracing for the sine electrode was simulated for 5 s, while for the fishbone and the snake it was simulated for 10 s, because some particles were temporarily trapped, and more time was needed for them to reach the end, in the latter two cases. The particle-tracing simulations are shown in Fig. 13 and the supporting videos in Supplementary items 4–6.

The amount of the collected particles was determined by how many of them reached the middle third of the channel outlet, shown by the blue area in Fig. 13. Particles that either did not reach the outlet or exited through the side thirds were considered uncollected. The efficiency of the particle collection calculated by the following formula:

$$\text{collection efficiency} = \frac{\text{collected particles}}{\text{all particle}} \times 100\%, \quad (14)$$

During the simulations, some particles were removed because of the many wall interactions. These particles are not counted in this formula. The fishbone electrode (Supplementary item 4) collected one third of particles (33 % efficiency), which is equal to what it would have been collected without turning it on. Based on this simulation, this design is not suitable for effective particle collection. In the snake electrode (Supplementary item 5) simulations, 47 particles were collected, 31 escaped and 12 “removed” (permanently trapped). This resulted in a collection efficiency of 60.3 %. For the sine design (Supplementary item 6), we got better results: 69 particles were collected, 3 were not and 14 removed, leading to an efficiency of 95.8 %. (The removed fraction was



Video 6. Particle collecting experiments for 1 μm particles with the sine electrode at a flow rate of 0.1 $\mu\text{L}/\text{min}$. A video clip is available online. Supplementary material related to this article can be found online at [doi:10.1016/j.colsurfa.2026.140045](https://doi.org/10.1016/j.colsurfa.2026.140045).

attributed to the improper modeling of the interaction between the electrode surface and the beads; hence, it was not counted when calculating the efficiency, therefore, these results are considered only upper estimates.)

3.2. Experiments

During the real-life experiments, we first worked with 1 μm polystyrene microbeads (Polybead Microspheres 1.00 μm , Polysciences). The frequency and the amplitude were chosen 500 kHz and 7 V, respectively. The voltage was selected according to the lowest value above which the efficiency no longer increases noticeably, and the particle collection efficiency was measured under different fluid flowrates, from 0.005 $\mu\text{L}/\text{min}$ to 0.1 $\mu\text{L}/\text{min}$. Snapshots of the experiments are shown in Fig. 14.

In the simulation, only a few particles were trapped between the electrodes, while in the experiments nearly all particles were trapped. One possible explanation is particle adhesion to the glass surface and the electrodes in the experimental setup. As a result, a stronger drag force is required to move the particles that are already attached.

In the post-electrode section, the particle flow into the middle of the channel only began after the inter-electrode space got “saturated”, and the particles started pushing off each other, which usually took a few minutes. The saturation is defined qualitatively and may depend on particle concentration and imaging conditions. However, this does not cause any problem in terms of particle collection: after the full saturation of the electrodes, particles begin to flow toward the outlet with high efficiency. Prior to saturation, particle collection is less effective. We consider the inter-electrode space saturated when the surface is no longer visible due to the high particle concentration. The saturation time can be shortened by increasing the particle concentration. These two

quantities are inversely proportional to each other.

In the post-electrode region of the channel the only force acting in the particle is the drag force from laminar flow, therefore the particles exhibit no lateral movement. This region was divided into three equal areas parallel to the channel walls, corresponding to the three outlets. The particles in the areas will leave the channel at their respective output. We counted the particles in each area, with the middle area was designated for particle collection and the side areas for waste. The collection efficiency was then calculated using Eq. 14. The experimental results were consistent with the simulations.

The fishbone structure had very narrow trapping range in both cases, hence the collection efficiency was below 50 %. The snake electrode achieved an efficiency of approximately 70 %, which is slightly higher than predicted by the simulations; however, this performance was limited to flow rates below 1 $\mu\text{L}/\text{min}$. Increasing flow rates, limiting the time the electrodes interact with the beads more and more, gradually weakens the collection efficiency. In addition, because the snake electrode’s trapping sites are not arranged continuously, particles can easily escape the electric field.

In contrast, the sine electrode has a continuous structure, making it more difficult for the particles to escape. At very low flow rates, around 0.005 $\mu\text{L}/\text{min}$, we observe enhanced permanent particle trapping, which reduces the collection efficiency to only approximately 50 %, however, as the flow rate increases, the collection efficiency also increases. At around 0.015 $\mu\text{L}/\text{min}$, the sine electrode already outperforms the snake electrode, reaching an efficiency of 87 %. It should be noted that above 85 %, our method underestimates the efficiency, due to an increasing probability of particle overlaps, so the true efficiency could reach 90 % or higher. By further increasing the flow rate, the sine electrode continues to improve, and at 0.1 $\mu\text{L}/\text{min}$ it achieves a collection efficiency exceeding 95 %. (Particle collection for the sine electrode shown in Supplementary items 7–9.) The comparison of collection efficiencies from simulations and experimental measurements are shown in Table 2.

In view of potential biological applications, it was important to clarify whether these electrode arrangements are also able to capture smaller particles, as well. Therefore, in the subsequent experiments 0.1 μm diameter latex microbeads were used, stained by fluorescent dyes (Red Fluorescent Carboxylated PS Latex, 0.10 μm 2.5 %, MagSphere), and excited at a wavelength of 532 nm, to ensure visibility. The

Table 2

Comparison of efficiencies from simulations and experiments.

	Maximum simulation efficiency	Maximum experimental efficiency
Fishbone	33 %	33 %
Snake	60.3 %	70 %
Sine	95.8 %	97 %

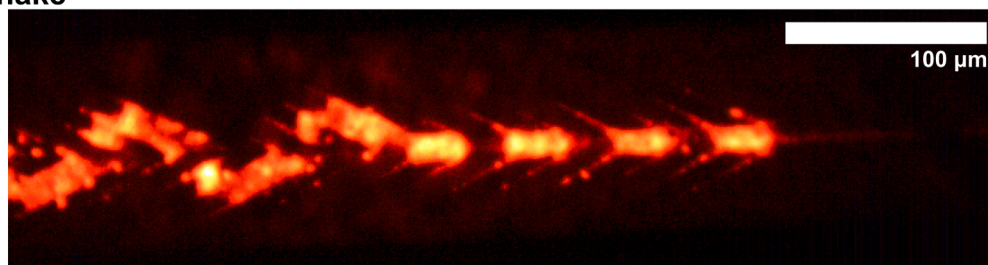
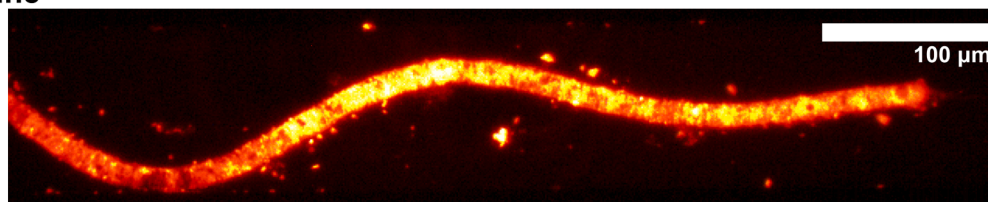
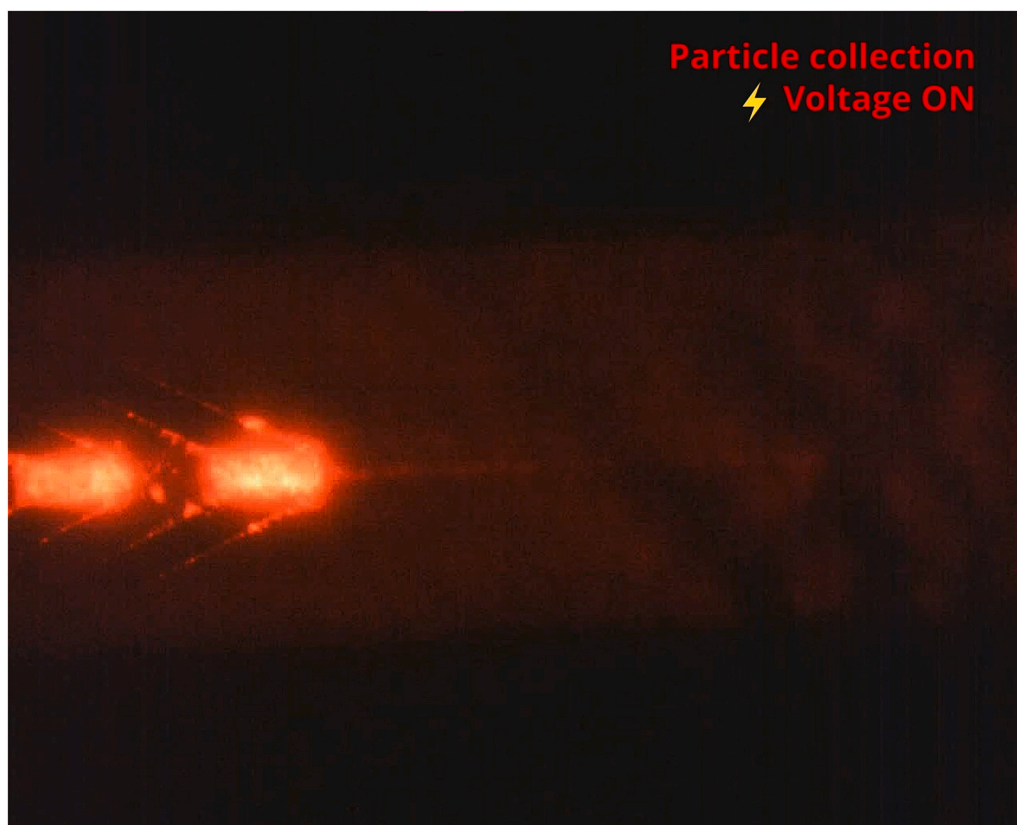
Snake**Sine**

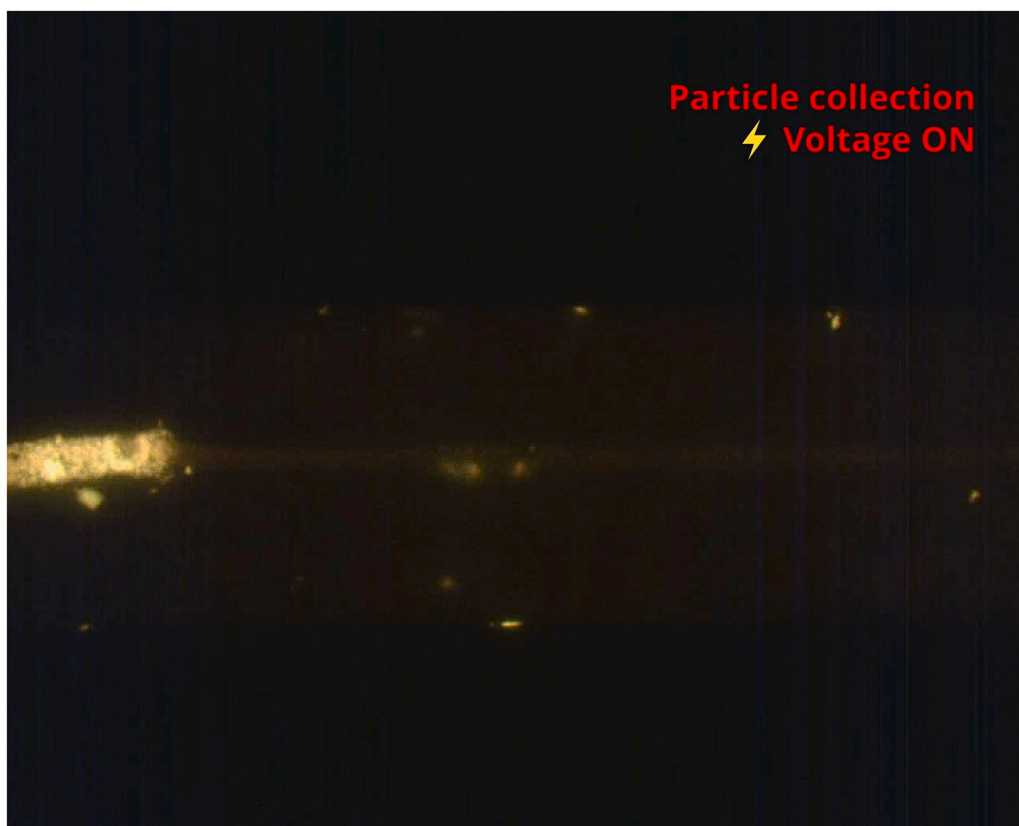
Fig. 15. Snake (up) and sine (down) electrodes in operation, 10x magnification. The 0.1 μm red fluorescing particles were collected at 500 kHz, 15 V and 1 $\mu\text{L}/\text{min}$ fluid flow rate. The particle collections are shown in the supporting videos ([Supplementary items 10 and 11](#)).

frequency of the electric field was set to 500 kHz, the voltage to 15 V, and the fluid flow rate to 1 $\mu\text{L}/\text{min}$. Experiments were conducted using the snake ([Supplementary item 10](#)) and the sine ([Supplementary item 11](#)) electrodes, with the results shown in [Fig. 15](#). Because of their smaller size, these particles require a somewhat larger voltage (according to the balance of the DEP- and drag forces) for effective collection. While the 1-

μm particles are effectively collected by 7 V, collection of the smaller particles was minimal at this voltage, and the effective collection began only above 10 V. The simulations indicate that significantly higher voltages are required to achieve the same DEP force as was effective for the larger particles. When, however, we calculate the speed component of the particles of different size moving toward the electrodes (v_e) due to



Video 7. Particle collecting experiments for 0.10 μm fluorescents particles with the snake electrode at a flow rate of 1 $\mu\text{L}/\text{min}$. A video clip is available online. Supplementary material related to this article can be found online at [doi:10.1016/j.colsurfa.2026.140045](https://doi.org/10.1016/j.colsurfa.2026.140045).



Video 8. Particle collecting experiments for $0.10\ \mu\text{m}$ fluorescent particles with the sine electrode at a flow rate of $1\ \mu\text{L}/\text{min}$. A video clip is available online. Supplementary material related to this article can be found online at [doi:10.1016/j.colsurfa.2026.140045](https://doi.org/10.1016/j.colsurfa.2026.140045).

the DEP force, substantially lower voltages are sufficient to achieve particle collection for smaller beads, due to the lower friction force in this case, as we pointed out under [Section 3.1.1](#). All-in-all, the practical voltage requirements are governed by the balance between DEP force and hydrodynamic drag (particle velocity), rather than DEP force magnitude alone.

Moreover, operating at lower voltages is preferable, as it reduces Joule heating and other electrothermal effects. Similar to larger particles, the trapped beads began moving toward the outlet only after the electrodes became saturated. However, because of their smaller size, a much higher number of particles is required for saturation in this case. This was compensated with larger concentration and higher flow rate. Of course, higher concentration are not always achievable. To get around this problem, the electric field can be turned on and off by a trigger signal (Uniblitz model VMM-T1 shutter driver/timer). When the electric field was switched on ('on') the particles were trapped, and when the field was switched off ('off'), the electrodes released them. By optimizing the flow rate and the 'on' and 'off' times, the particles could be released at a frequency that precisely matched their flow speed to be trapped by the next electrode. This method was also useful to prevent permanent trapping. [The Supplementary items 10 and 11](#) videos show the electrodes during particle collection. The release of particles upon switching off the voltage is presented solely to demonstrate the trapping effect. After reapplying the voltage, the capturing of the particles by the DEP field begins, and after a short time the particle collection to the outlet also starts.

We note that at lower frequencies it is important to monitor the integrity of the electrodes, because as the frequency decreases, instantaneous erosion of the electrodes might occur, rendering the system unusable. In distilled water medium, e.g., the erosion begins below $50\ \text{kHz}$ at $5\ \text{V}$, and this threshold varies depending on the ionic

composition of the medium. For example, in $300\ \text{mmol/L}$ sucrose solution, the erosion begins below $500\ \text{Hz}$, but in $10\ \text{mM NaCl}$ solution this value is increased to $100\ \text{kHz}$. The erosion could be reduced by using a protective coating, for example poly-HEMA, too [\[8\]](#).

In conclusion, we could establish that particle collection with these systems works efficiently. One of our future plans is to optimize the experimental conditions for smaller particle concentrations, by switching the electric field on and off. By fine-tuning the on- and off- intervals, we can get around permanent trapping of the particles. At optimal timing, the electric field switches off when the particles reach the next particle-capturing area. This on-off technique can enhance the efficiency of the snake electrode, which traps the particles at discrete steps, not continuously like the sine electrode. Another potential method to lower permanent trapping is using a protective layer that prevents sticking of the particles to the surface. This layer also can protect the electrodes from damage at lower frequencies.

Having optimized particle collection, our future goal is to apply this strategy for the collection and separation of submicron biological particles, such as extracellular vesicles.

4. Conclusion

We simulated and tested three different electrode designs for positive dielectrophoretic particle-collecting ability of $1\ \mu\text{m}$ polystyrene particles. The three designs were named fishbone, snake and sine. The DEP-force calculation showed that the best designs for the particle collection are the snake and the sine structures, which can collect particles in the full width of the channel, while the fishbone has a narrow range. According to the particle-tracing simulations, the fishbone structure collected only $33\ \%$ of the particles, the snake's efficiency was $60.3\ \%$, and finally the efficiency of the sine design was $95.8\ \%$. The

experimental measurements showed similar results, along with that the snake electrode showed somewhat higher efficiency than in the simulation (around 70 %), while the sine electrode outperformed the other structures, reaching more than 95 % efficiency. We have shown that these designs are able to efficiently collect smaller (0.1 μm) particles, as well, implying important future applications in medical diagnostics, concerning, e.g., extracellular vesicles that are of the same size range.

CRedit authorship contribution statement

Máté Farkas: Writing – original draft, Visualization, Data curation, Investigation, Validation, Formal analysis. **Sándor Valkai:** Conceptualization, Writing – review and editing, Methodology. **András Dér:** Conceptualization, Funding acquisition, Resources, Supervision, Writing – review and editing, Methodology, Validation.

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Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Appendix A. Supporting information

Supplementary data associated with this article can be found in the online version at [doi:10.1016/j.colsurfa.2026.140045](https://doi.org/10.1016/j.colsurfa.2026.140045).

Data availability

Data will be made available on request.

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