Active Posts in Deterministic Lateral Displacement Devices

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Using electrically connected metal-coated posts in a deterministic lateral displacement (DLD) device and applying electric fields, electrokinetics is used to tune separations, significantly decrease the critical size for separation, and increase the dynamic range with switching times on the order of seconds. The strength of DLD stems from its binary behavior. To first approximation, particles move in one out of two trajectories based on their effective size. For particles that are close to the threshold size, a small external force is sufficient to nudge the particles from one trajectory to another. The devices consist of arrays of cylindrical metal-coated SU-8 posts connected by an underlying metal layer. This allows the application of voltages at the post surfaces and the generation of electric field gradients between neighboring posts, causing polarizable particles to experience a dielectrophoretic (DEP) force. This force, which depends on the volume and polarizability of the particle, can be made sufficient to push particles from one trajectory into another. In this way, the critical size in a device, normally fixed by the geometry, can be tuned. What’s more, adding DEP in this way allows for the simultaneous creation of multiple size fractions.

1. Introduction

In many types of miniaturized integrated fluidic systems, particle sorting is essential for sample preparation and analytical fractionation. Microfluidics simplifies the process considerably compared to standard methods such as fluorescence-activated cell sorting (FACS) and magnetic-activated cell sorting (MACS). For high-throughput applications, acoustophoresis and inertial focusing have been developed. However, they provide limited tunability of the separation functionality requires additional refinement. Beech and Tegenfeldt used the deformability of PDMS to change the DLD device geometry by pulling on it and thereby change the critical size. It requires the application of electric fields but most relevant to the current work, Beech et al. used innate, long-range electrostatic forces inside DLD devices, modulated via buffer ionic strength, to tune separations and achieve a large reduction of the critical size. Zeming et al. recently used this approach to detect the binding of proteins to the surface of micrometer-sized beads. In both of the above cases, nanometer-sized particles are separated in devices with micrometer-sized features. As the authors argue, this approach delivers powerful separations in devices that are easy to fabricate (large features) and easy to run (low pressures). Most relevant to the current work, Beech et al. used electric fields to modify the behavior of polystyrene microspheres in DLD devices. This approach also enables the tuning of devices and the decrease of the critical size. It requires the application of electric fields but has the advantage that tuning can be done quickly and reversibly without the need to change the buffer. These devices were simple to fabricate and use, requiring only electrodes added at the inlets and outlets of devices. The insulating properties of PDMS generate insulator-based dielectrophoretic (DEP) forces that tune the separations in DLD devices. This approach has now been refined by Ho et al. (submitted for publication) who...
have used buffer conditions and applied voltages and frequencies to separate microspheres, bacteria, and yeast based not only on size, but on surface charge and other dielectric properties. However, generating high electric fields in devices using this approach requires high applied voltages (typically hundreds to thousands of volts), which can be problematic from a practical point of view, and it can be difficult to do this at high frequencies without specialized equipment.

Here, we present a new approach to combine DEP and DLD by fabricating post arrays that also serve as electrodes. These three-dimensional electrodes have been used previously for an impedance flow cytometer\[8a\] and for parallelized electrorotation in a microfluidic channel.\[8b\] With our new design, we are able to generate similar field gradients as in our previous work\[6\] using two orders of magnitude lower applied voltages due to the proximity of the electrodes (Figures 1 and 2). This allows us to work with frequencies up to tens of megahertz using standard signal generators. Using this approach, we reversibly decrease the critical particle size in a device from 6 to 0.25 µm, a factor of 24, equivalent to increasing the dynamic range from 1 to 24. The switching times for adjustment of critical sizes within the demonstrated range are on the order of seconds.

![Figure 1. Device principle. A) Particles smaller than the critical size in a DLD device move in the zigzag mode, following the average fluid flow direction (both solid and dashed red lines) while those larger than the critical size flow in the displacement mode (blue line). B) If the posts are active electrodes, then electric fields can be generated between the rows of posts in such a way that DEP forces cause particles to transition to the displacement mode (the inset shows an illustration of the forces). C) Using this principle, devices can be designed such that particles with the same size but different DEP mobilities can be separated and subsequently collected (solid red line).](image1)

![Figure 2. A) Schematic of device design showing how the electrodes are coupled. B) Photograph of a finished device showing inlet and outlet reservoirs and wires for fluid and electronic interfacing. C) Colorized SEM micrograph of the active post region. The posts that constitute the DLD array are coated with metal (shown in red and green) and connected via buried metal lines. The red and green colorized electrodes are connected as depicted in (A) to have opposite polarity.](image2)
2. Theory

2.1. Deterministic Lateral Displacement

The critical particle diameter \( D^C \) in a DLD array with gap \( G \) and period \( N \) can be estimated using the empirical formula\(^9\)

\[
D^C = 1.4 \cdot G \cdot N^{-0.48}
\]  

(1)

As shown in Figure S2 in the Supporting Information, the measured gap in our device, \( G = 11.10 \pm 0.14 \) \( \mu \)m. With the period designed to be \( N = 10 \), the critical particle diameter is expected to be \( D^C = 5.15 \pm 0.06 \) \( \mu \)m and is measured experimentally (Section 3 and Figure 3) to be \( \approx 6 \) \( \mu \)m.

This means that particles with diameters smaller than 6 \( \mu \)m are expected to follow the fluid flow direction in the device (commonly referred to as the zigzag mode) and those larger than 6 \( \mu \)m will follow the geometry of the array (bumping or displacement mode). Here, we will refer to the two modes as zigzag and displacement. Further studies have refined our understanding of DLD mechanisms\(^{10,11}\) and have shown the existence of a mode that lies between the zigzag and displacement modes and that has its origins in anisotropic permeability\(^{10,12}\) and broken flow symmetry.\(^{13}\) We observe this mixed mode in our measurements (Figure 3), and observe how particles switch from zigzag to displacement via the mixed mode.

2.2. Dielectrophoresis

Dielectrophoresis provides specificity based on both the volume and the dielectric properties of the particles.

The resistance to the formation of an electric field, the complex permittivity \( \tilde{\epsilon} \), inside a material upon the application of an external alternating field is a function of frequency, \( f \), and is given by the following where \( \epsilon \) is the permittivity (in a static field) and \( \sigma \) is the conductivity

\[
\tilde{\epsilon} = \epsilon - j \frac{\sigma}{2\pi f}
\]  

(2)

The time-averaged force \( \langle F_{\text{DEP}} \rangle \) on a spherical particle of radius \( r \) and complex permittivity \( \tilde{\epsilon}_p \) suspended in a medium with complex permittivity \( \tilde{\epsilon}_m \) and a sinusoidal electric field with an amplitude of \( E \) is given by

\[
\langle F_{\text{DEP}} \rangle = \pi r^3 \text{Re}(f_{\text{CM}}) \nabla E^2
\]  

(3)

The force depends on the contrast in dielectric properties between the particle and the surrounding medium as expressed by the real part of the Clausius–Mossotti factor \( f_{\text{CM}} \) (Equation (4)). This factor also carries the frequency dependence of the system. By tuning the frequency, different aspects of the particles are targeted for separation. As the frequency goes to zero (typically surface), conductivity dominates. As the frequency goes to infinity (typically bulk), permittivity dominates. Thus, for low frequencies, the surface of particles is probed whereas at high frequencies the bulk is probed. As a rule of thumb, positive DEP is the result at low frequencies and low medium conductivities. At high frequencies, the permittivity contrast between the particle and the medium dominates, and with the high permittivity of water, the resulting DEP is negative

\[
\text{Re} [f_{\text{CM}}] = \frac{\tilde{\epsilon}_p - \tilde{\epsilon}_m}{\tilde{\epsilon}_p + 2\tilde{\epsilon}_m}
\]  

(4)

A thorough treatment of DEP can be found in Pethig’s textbook.\(^{14}\)

2.3. Active Posts

As mentioned above, the major advantage of placing electrodes inside devices is that the maximum field gradient can be generated exactly where it has the greatest effect on the trajectories of the particles, using minimal applied voltage. While a similar effect could be achieved using a combination of 2D electrodes and 3D fluidic structures (for example, metal electrodes on a surface with PDMS structures on top), 3D electrodes have the major advantage that they provide a homogeneous electric field distribution over the complete depth of the device.\(^{8a–d}\) This homogeneous distribution results in
forces that are invariant in the depth direction and the method is therefore also insensitive to buoyancy effects.

2.4. Combining DLD and DEP Using Active Posts

In a DLD device, steric interactions between particles and posts cause particles to cross streamlines and to follow distinct trajectories through the device. As will be shown in Section 3, a DEP force can be created that, together with the steric interaction, can be sufficient to force a transition from the zigzag to the displacement mode. As described above, the DEP force is a function of the volume of the particle, the polarizabilities of the fluid and particles at the specific frequency, and the gradient of the electric field squared. In contrast to previous work by Beech et al.\[6\] with electrodeless DEP in DLD devices, placing the electrodes onto the surface of the posts allows high field gradients to be achieved over a large range of frequencies (0–10 MHz with our current function generator). The significant advantages of combining DEP and DLD in this way are that we can tune the critical size in a device (that otherwise only has one critical size) by simply turning the knobs on our function generator. As will be shown below, we can decrease the critical size by a factor of 24 and do so with a response time of seconds. This means that we can separate 4 \( \mu \)m particles from 6 \( \mu \)m particles, or, 250 nm particles from 500 nm in the same device. What’s more, the gaps between the posts in the device are 10 \( \mu \)m, making clogging negligible. We have run separations continuously for 10 h without clogging. Also, while the focus in this report is on size-based separation, the dependence of the DEP force on the polarizability of particles means that we also have a handle on separations by dielectric properties which can further aid us in identifying different types of cells or microorganisms as well as in differentiating cells in different states, e.g., in live/dead assays.

3. Results and Discussion

Figure 4 shows an example of the switching of modes (zigzag to displacement) in our device. In Figure 4A, 4.3 \( \mu \)m microspheres, being smaller than the 6 \( \mu \)m critical size in the device, move in the zigzag mode. Upon the application of 2 Vpp at 10 MHz (Figure 4B), the microspheres switch to the displacement mode. In Figure 4C, the trajectories of single microspheres are shown. In the bottom panel at 0 Vpp, the microsphere is in the zigzag mode. At 1.5 Vpp, the period in the zigzagging motion increases leading to greater displacement. At 2 Vpp, the microspheres are in full displacement mode. The change in the trajectories leads to the microspheres leaving the device at increasing outlet channel number as the applied voltage is increased. The position of each particle is established as shown in Figure 4E and the distributions plotted as shown in Figure 4D. In this way, the critical size in the device is tuned. Figure 3 shows outlet positions for seven different microparticle sizes and demonstrates how the critical size in the device changes as the applied voltage is increased from 0 to 2 V at 10 MHz.

We tested the response of the device to voltage scans at frequencies ranging from 1 Hz to 10 MHz. We find little to no separation below 100 kHz and also that at frequencies of 10 kHz and lower, at voltages above 3 Vpp, there is a tendency for microspheres to become trapped at the high field regions between the posts (see Figures S5 and S6 in the Supporting Information). We believe that this is likely due to positive dielectrophoresis or possibly AC electroosmosis although further experiments and simulations are needed to understand the mechanism in detail. We expect the optimal frequencies for tunable separation to depend on the types of particles being separated and frequency scans may well be the best approach to distinguish bioparticles with different polarizabilities. While tunable particle trapping by frequency scanning in these devices

![Figure 4](https://example.com/figure4.png)

**Figure 4.** Typical separation results. A) 4.3 \( \mu \)m microspheres are smaller than the native critical diameter \((D_C \approx 6 \mu m)\) in the device and therefore move in the zigzag mode. The image shows the entire separation area of the device. B) The same beads move in the displacement mode when 2 V @ 10 MHz is applied. Half way along the device, the microspheres hit the sidewall which leads to the zigzagging trajectory. C) Trajectories of individual 2.1 \( \mu \)m microspheres as a function of applied voltage (10 MHz). D) Measured outlet channel numbers as a function of applied voltage for the same microspheres shown in (C). E) Analysis is performed by measuring the exit channel number for every microsphere. Red arrows show microspheres flowing out of the device and the blue arrow shows a stuck microsphere that is not counted.
may, in its own right, be interesting for a range of applications, here we have focused on studies of the behavior of the device at a single frequency. For the polystyrene microspheres we study here, all with the same polarizabilities, we find 10 MHz to be optimal for separation (see Figure S5 in the Supporting Information) and chose to keep the frequency fixed at 10 MHz and study the voltage dependence.

Figure 3 shows the effect of applied fields on the separation behavior of our device for a range of particle diameters. The green curve shows the device functioning without applied electric field. Microspheres with diameters below ≈5 μm are in the zigzag mode and experience zero displacement. Microspheres with diameters above ≈7 μm are fully displaced. Microspheres with diameters of ≈6 μm are in the mixed mode and are only partially displaced. The critical size in this device, at zero applied field, can be approximated to ≈6 μm. When 1 Vpp @ 10 MHz is applied to the device, the effective critical size in the device decreases to ≈4.5 μm as shown by the red curve. At 2 Vpp, the critical size in the device decreases further to ≈1.5 μm.

One obvious question here is how small we can make the critical size. Figure 5 shows the tuned separation of 0.25, 0.5, and 0.8 μm microspheres at 10 MHz. At 0 Vpp, all microspheres are in the zigzag mode. At 3.5 Vpp, the 0.8 μm microspheres move mostly in the displacement mode and become separated. At 5.25 Vpp, the 0.8 μm microspheres are fully in the displacement mode, the 0.5 μm are in the mixed mode, and the 0.25 μm still in the zigzag mode. Here, the three microsphere populations are fully separated. This is particularly interesting because the device has only one critical size and if used in the traditional way, without applied fields, can usually only generate two fractions reliably. The mixed mode described above is difficult to predict and is most often seen as a factor that reduces device functionality and resolution. Here, we can reproducibly put particles in the mixed mode and create more than the expected 2 modes. At 7 Vpp, we are back to 2 fractions, one containing 0.25 μm microspheres and the other containing 0.5 μm and 0.8 μm microspheres.

The trajectories (measured via outlet position) of eight microsphere populations as a function of applied voltage at 10 MHz are shown in Figure 6. The same data are split into two separate plots (Figure 6B,C) for clarity. As the voltage is increased, each microsphere population, in decreasing size order, makes the transition from the zigzag to the displacement modes via intermediate, mixed modes. In Figure 6A as mentioned above, the 7.81 μm microspheres are displaced even at 0 Vpp applied voltage and 6.1 μm are in the mixed mode half way between zigzag and displacement modes indicating that the geometry-based (zero-field) ΔR in the device is ≈6 μm. In Figure 6A,C, the 0.25 μm particles enter the device with a slight offset (channel 9 rather than 8) which was due to a small clog in the inlet. The transitions are described well by error functions, shown fitted to the data. The error bars increase with decreasing particle size, consistent with diffusional broadening of the particle stream. Voltages can be found at which all particles are either displaced, nondisplaced, or somewhere in between and this can be changed with a switching time of seconds (see Section S8 in the Supporting Information). This is the fundamental principle of tuning using our method.

As the particles pass between the posts, there is a force balance between the viscous drag force and the DEP force that determines whether there is a transition from zigzag mode to displacement mode (Figure 7; Figure S7, Supporting Information). In Section S6 in the Supporting Information, we derive a scaling expression for the crossover voltage, \( V_C \):

\[
V_C^2 \propto \frac{R_C - R_{\text{particle}}}{R_{\text{particle}}} \frac{1}{\varepsilon_m \Re(f_{CM})} \Delta P \frac{\mu^4}{L}
\]  

Interestingly, the crossover voltage is independent of viscosity so that for a given device at a given applied pressure, \( \Delta P \), the only important factors are the particle size, \( R_{\text{particle}} \), in relation to the critical size, \( R_C \), the dielectric properties of the medium and of the particle.

We verify the scaling relationship by two plots. First, we plot the crossover voltage versus the square root of the applied pressure mean microspheres and the other containing 0.5 μm and 0.8 μm microspheres.

![Figure 5](image-url)  
Figure 5. Tuned separation of 0.25, 0.5, and 0.8 μm microspheres at 10 MHz and at A) 0 Vpp, B) 3.5 Vpp, C) 5.25 Vpp, and D) 7 Vpp. 0.25 and 0.5 μm microspheres are green fluorescent and 0.8 μm are blue fluorescent and were imaged with separate filter sets. The images were extracted from a movie, colorized and recombined to show all particles simultaneously. See the Supporting Information for the original movie.
in Figure 8A. Here, we use two criteria for the crossover. For each pressure, the voltage was tuned until the 4.3 µm microspheres exited at the mean exit channel or at the threshold of maximum displacement. $V_C$ is clearly linearly proportional to the square root of the applied pressure in both cases. Second, we investigate the relationship between the crossover voltage and the particle size (Figure 8B). The crossover voltage is extracted from the fitted error functions in Figure 6 as the voltage at which the particles exit at the mean exit channel, and the critical size is taken from the error function fit in Figure 3 at zero applied voltage to be $2R_C = 6.08$ µm. Also, here we can demonstrate a linear relationship, $V_C$ appears to be linearly proportional to $((R_C - R_{\text{particle}})/R_{\text{particle}})^{0.5}$, further supporting the validity of Equation (5).

The results of our simulations indicate that negative DEP is the force responsible for altering the trajectories of the particles. Figure 9B shows that there is good qualitative agreement between our experimental results and our simulations that take into account only forces due to drag, DEP, and steric interactions with the posts. Despite simplifications in the simulations (as explained above and in Section S4 in the Supporting Information) leading to an underestimation of the critical size, the transition from zigzag to displacement mode is captured faithfully. Interestingly, the mixed modes are not observed in the simulations. This is probably due to the fact that no anisotropy in the flow is present in the simulation. Figure 9A shows that greater negative DEP forces lead to greater displacements.

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**Figure 6.** Outlet positions as a function of applied voltage (at 10 MHz) for A) 0.25, 0.51, 0.80, 2.08, 2.90, 4.30, 6.10, and 7.81 µm microspheres. Particles exit at channel 8 if they experience no displacement and at 14 if they experience maximal displacement. The dashed lines show fitted error functions (see the Supporting Information for more information on fitting) except for the 7.81 µm microspheres which is a spline. Error bars show one standard deviation. (B) and (C) show the same data with expanded x-axis for clarity.

**Figure 7.** Illustration of the lateral components of the viscous drag force and DEP force. The viscous drag force acts to keep the particle in the zigzag mode whereas the DEP force pushes the particle into the displacement mode.
As Davis discusses in his thesis, the maximum particle size that can be handled is limited to the smallest gap size in the device, which ultimately limits dynamic range to between 3 and 5. Cascaded devices have channels that remove large particles which could be a problem at excessive applied voltages. Alternatively, we can also consider the lower limit for $D_C$ as the critical size by the same factor as shown here (24) would be able to decrease the critical size further although heating is occurring under the conditions used for separation but we can at least conclude that we are working far below the limits for excessive heating and that we have strategies (such as cooling) that could be used where heating could be a problem, such as for bioseparations requiring high ionic strengths. We also observed some hysteresis after boiling, with the threshold decreasing after an initial boiling event. This, we believe, could be due to damage to the device although full breakdown of the electrodes and SU8 structures in the device occurred first at voltages well above 100 Vpp.

4. Conclusion

We have successfully demonstrated proof of principle of a DLD device combined with electrodes directly integrated onto the posts. The short interelectrode distances allow us to reach high electric field strengths, on the order of nominally 1 MV m$^{-1}$, at high frequencies with small enough voltages to ensure simple and safe operation. We have shown that we can change the critical size in our device from $D_C = 6$ µm to 250 nm, a decrease by a factor of 24, which constitutes an increase in dynamic range by the same factor. We predict that at higher applied voltages, we will be able to decrease the critical size further although heating could be a problem at excessive applied voltages. Alternatively, devices with a smaller critical size of $D_C = 2$ µm would be within the capabilities of the fabrication method we have used here. Applying electric fields to these devices and decreasing the critical size by the same factor as shown here (24) would allow us to separate particles in the sub 100 nm diameter range. Switching times in our device depend on flow rates, but at the conditions presented here, they are in the range of several seconds. This means that we can immediately visualize the response to a change in voltage and tune separations in real time.

The dynamic range of a DLD device can be defined as the ratio of the largest to smallest critical diameters that can be achieved. Increasing this range is often done using so-called chirped arrays, i.e., multiple DLD array in sequence. However, as Davis discusses in his thesis, the maximum particle size that can be handled is limited to the smallest gap size in the device, which ultimately limits dynamic range to between 3 and 5. Cascaded devices have channels that remove large particles between the cascaded arrays and in these dynamic ranges of 20 can be achieved, but at the cost of increased complexity of both design and performing separations.

Because our device has only one critical size ($D_C = 6$ µm), it has a dynamic range of one, when not using the active post capabilities. However, since we are able to displace 0.25 µm particles in our device by applying electric fields, we have increased the dynamic range to ≈24. By combining our active post approach with the chirped and cascaded device approach, we believe we could achieve dynamic ranges greater than 100.

Lastly, we can also consider the lower limit for $D_C$ as the applied voltage is increased. With our current setup, we were unable to apply higher voltages at 10 MHz than those required to displace 250 nm particles. However, at frequencies below 1 MHz, we were able to test the maximum voltage that the device could sustain. At ≈40 Vpp, rapid bubble formation was observed. The threshold for bubble formation was observed to decrease with increasing ionic strength ($>30$ Vpp for 1500 mS m$^{-1}$ KCl compared to $≈40$ Vpp for milliQ at $≈0$ mS m$^{-1}$). Bubbles also formed at lower voltages for lower frequencies ($≈30$ Vpp @ 1 KHz compared to $≈40$ Vpp @ 1 MHz). Higher flow rates also decreased the threshold for bubble formation. At 25 mS m$^{-1}$ KCl, 40 Vpp @ 5 MHz bubbles formed at 77 mBar but not at 500 mBar. Lastly, cooling the device with ice also decreased the threshold slightly. Taken together, these results indicate that bubble formation is probably due to joule heating where the power density is proportional to the electric field squared and to the conductivity. Further studies are required to ascertain exactly how much heating is occurring under the conditions used for separation but we can at least conclude that we are working far below the limits for excessive heating and that we have strategies (such as cooling) that could be used where heating could be a problem, such as for bioseparations requiring high ionic strengths.

Figure 8. Crossover voltages as a function of applied pressure and particle size. A) Red crosses show crossover voltages (mean exit channel number is 11) for 4.30 µm microspheres as a function of the square root of the applied pressure. The black crosses show the voltage at which maximum displacement (mean exit channel number is 14) is first reached. R-square of the fit is 0.9979. B) Crossover voltage as a function of $((R_c - R_{part})/R_{part})^{3/2}$. Here, the R-square of the fit is 0.9826. In all cases, the error bars show one standard deviation and the lines show a linear fit with 95% confidence intervals.
time. At high flow rates or applied voltages, the switching times will be shorter. To summarize, we change what is essentially a bimodal separation technique into a tunable multimodal separation tool. Further studies will focus on the application of the method to biological particle systems.

5. Experimental Section

Device Fabrication: The detailed fabrication process is described in Figure S1 in the Supporting Information. A metal layer (Ti/Pt/Ti 20 nm/200 nm/20 nm) was first sputtered onto a plane glass substrate and patterned by ion beam etching. The metal lines were then covered with an insulating SiO2 layer (300 nm) and via’s to connect the electrodes were etched by reactive ion etching. To create the active electrode structures, SU-8 pillars were patterned (15 µm thick) and subsequently coated by metal sputtering (Ti/Pt 20 nm/200 nm). The metal deposited on top of the pillars and on the planar substrate was removed by directional ion beam etching. This etching process preserved the metal-coated side walls of the electrodes.

A second SU-8 lithography step was used to define the fluidic channels and all other nonelectrically active device features. Channels were sealed with an 8 mikron thick PDMS slab using oxygen plasma and APTES. Before sealing, holes were punched through the slab for fluidic access. After sealing, silicone tubes were glued to the PDMS to function as fluid reservoirs and for interfacing with a pressure control unit. Wires were soldered to the device for electrical connection. The final device is shown in Figure 2B. Further device details, such as the measured dimensions of the final device, can be found in Figure S2 in the Supporting Information.

Experimental Setup, Method, and Materials: Flow in the devices was generated by applying overpressure in the range of 2–75 mBar to the device inlets using an MFCS-4C pressure controller (Fluiqent, Paris, France). Outlet reservoirs were kept at ambient pressure. A function generator (15 MHz function/arbitrary waveform generator, model 33120A, Hewlett Packard, Palo Alto, CA, USA) was used for the application of AC signals. The applied voltage signal was monitored with an oscilloscope (Hewlett Packard 54603B 60 MHz) with a 1x/10x probe (Kenwood PC-54, 600 Vpp). All images were captured through a microscope (Nikon Eclipse TE2000-U, Nikon Corporation, Tokyo, Japan), with an Andor Neo CMOS camera (Andor Technology, Belfast, Northern Ireland) using NIS Element software (NIX Element Advanced Research v4.51, Nikon). The conductivities of the media and the suspensions used in all experiments were measured using a B-771 LAQUAtwin Compact Conductivity Meter (Horiba Instruments).

Particle Counting and Analysis: The separation of particles was determined by manual counting at the end of the DLD array. The process is described in detail in Section S3 in the Supporting Information and in Figure 4. In short, the number of particles exiting the device through each of the 14 exit channels was established by manual inspection of recorded movies for each of the experimental conditions (varied voltage, frequency, and pressure) and for each of the particle sizes.

Simulations: In order to gain insight into the distribution of the electric field inside the device and to aid in the understanding of how devices work, numerical simulations were performed in COMSOL Multiphysics 5.3a. In brief, trajectories of particles of different sizes were simulated moving through a device under conditions similar to those in the experiments. COMSOL’s “laminar flow” module and “electric current” module were used to simulate both fluid flow velocity and electric fields, and the trajectories of particles of different sizes moving through these fields were then calculated using the “particle tracing for fluid flow” module. Full details are given in Section S4 in the Supporting Information, and results can be seen in Figure 9 above.
Supporting Information
Supporting Information is available from the Wiley Online Library or from the author.

Acknowledgements
The parts of this work carried out at NanoLund were supported by the BeyondSeq consortium (EU Horizon2020 project 634890), the evFOUNDRY consortium (EU Horizon2020 project 801367), and the Swedish Research Council (grant no. 2016-05739), and those carried out at EPFL by the Swiss National Science Foundation (205321_179086). The authors are thankful for support from the Centre for MicroNanotechnology at EPFL, where the chips were fabricated.

Conflict of Interest
The authors declare no conflict of interest.

Keywords
3D electrodes, deterministic lateral displacement, dielectrophoresis, particle sorting, tunable separation

Received: April 23, 2019
Revised: June 7, 2019
Published online: July 11, 2019

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