USWNet 2010 Conference

October 2-3, Groningen , The Netherlands

Abstracts & Scientific program

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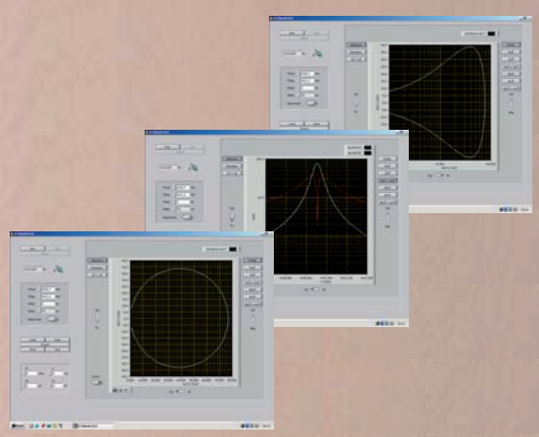

WELCOME TO USWnet 2010!

USWNet stands for 'Ultrasonic Standing Wave Network' and was initiated in 2003 in the UK bringing together a diversity of scientists and technologists with interests touching on particle and fluid movement by standing wave vibrations. The conference is aimed at fostering a deeper understanding of the phenomena and increasing their impact on research and commercial applications.

The 8th annual USWNet meeting is run as a satellite meeting preceding the 14th International Conference for Chemistry and Life Science, uTAS 2010, in Groningen www.microtas10.org.
The idea is to enable the global community of microfluidics to interact with the rapidly growing community of ultrasonic standing wave technology. Exciting developments are reported at an increased rate where the combinaton of microfluidics and cell and particle manipulation in microsystems by means of ultrasonic standing wave forces are key components for success.

It is my hope that by running the 8th USWNet conferece in close proximity with MicroTAS 2010 it will bring an added value to both communities and that new interdisciplinary initiatives and collaborations will be triggered.

Thomas Laurell, Chairman USWNet Conference 2010

SCIETNTIFIC PROGRAM, Saturday, October 2nd

All sessions will be held in "Red room".
Poster, coffe and lunch will be at "Blue Patio"

Time	Saturday October 2, 2010
10.00-12.00	USWnet Board meeting
12.00-13.30	Registration and Lunch
13.30-13.40	Conference opening - *Thomas Laurell*
13.40-14.20	Invited Lecture - *Minoru Seki*
Session I	
14.25-14.45	Sonotweezers: Towards 3D particle positioning *Peter Glynne-Jones, University of Southampton, UK*
14.45-15.05	Towards Integrated Acoustic Cell Trapping and Polymerase Chain Reaction *Brian Poe, University of Virginia, USA*
15.05-15.25	Frequency-shift-keying actuation with a damped transducer for ultrasonic particle aggregation in a multi-well chip *Mathias Ohlin, KTH, Sweden*
15.25-16.00	Break, posters & exhibition
Session II	
16.00-16.20	Quantifying acoustic streaming in large-particle acoustophoresis *Rune Barnkob, Technical University of Denmark, Denmark*
16.20-16.40	Circular flow of liquid induced by acoustic streaming *Dirk Möller, ETH Zurich, Switzerland*
16.40-17.00	Reduction of the influence of the sample fluid density over the resonance frequency of the acoustic modes in a microfluidic cavity *Muhammet Araz, Cornell University, USA*
17.00-17.10	Break
Session III	
17.10-17.30	Acoustophoretic contrast factors of living cells measured in acoustic fields calibrated in situ using polystyrene microbeads *Per Augustsson, Lund University, Sweden*
17.30-17.50	Ultrasonic extraction of tumour cells *Itziar Gonzalez, Agencia Consejo Superior de Invesetigaciones Científicas, Spain*
17.50-18.10	Numerical modelling of a quarter wave separator for the removal of lipid particles from pericardial suction blood *Giuliana Trippa, University of Oxford, UK*
19.30	**Dinner at Restaurant Feithhuis**

SCIETNTIFIC PROGRAM, Sunday, October 3rd

All sessions will be held in "Red room".
Poster, coffe and lunch will be at "Blue Patio"

Time	Sunday October 3, 2010
08.30-09.10	Invited lecture - *Joel Voldman*
Session IV	
09.20-09.40	Theoretical and numerical calculations for the time-averaged acoustic forces and torques experienced by a rigid elliptic cylinder in an ideal fluid *Jingtao Wang, ETH Zurich, Switzerland*
09.40-10.00	Hexagonal chamber for rotation of non-spherical particles *Thomas Schwarz, ETH Zurich, Switzerland*
10.00-10.20	Exploiting an USW for attenuated total reflection spectroscopy *Stefan Radel, Vienna University of Technology, Austria*
10.20-10.50	Break, posters & exhibition
Session V	
10.50-11.10	Acoustophoretic Preparation of Blood Components from Apheresis Product *Andreas Lenshof, Lund University, Sweden*
11.10-11.30	Development of a system for label-free somatic cell enumeration in raw milk using acoustophoresis *Carl Grenvall, Lund University, Sweden*
11.30-11.50	Piezoelectric ceramics or composites for micromanipulation in multilayer plastic resonators *Itziar Gonzalez, Agencia Consejo Superior de Investigaciones Científicas, Spain*
11.50-12.00	**Announcement of USWNet Conference 2011**
12.00-12.10	**Closing remarks**

Dinner, Saturday, October 2nd

The Feithhuis is located in a monumental building at the Martinikerkhof in Groningen. The interior is very spacious, exuding a relaxed and open atmosphere. The café, restaurant and terrace all have their own unique character.

Stadscafé-Restaurant 't Feithhuis
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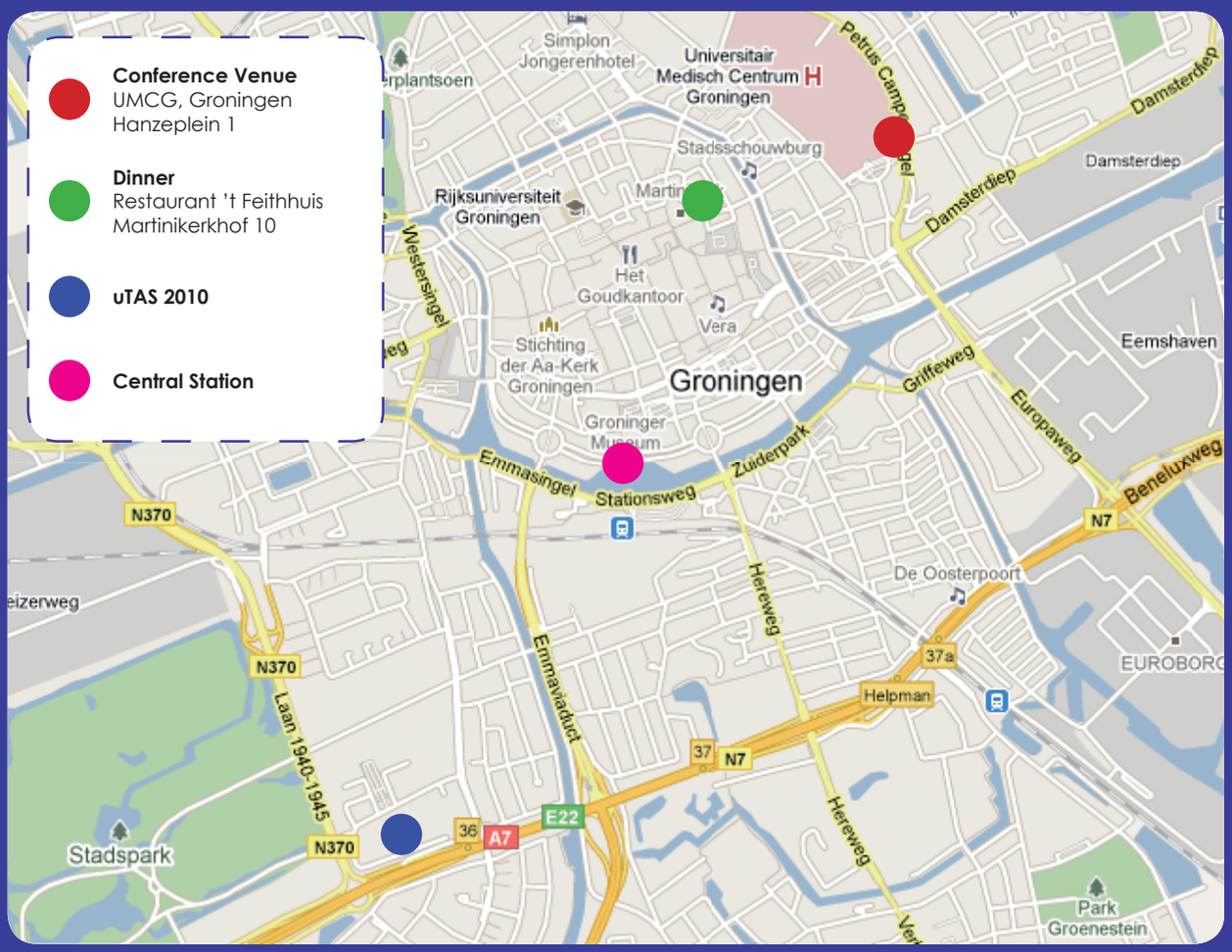
Map of Groningen



Floor plan

umcg University Medical Center Groningen

First Floor
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Hanzeplein 1, Groningen

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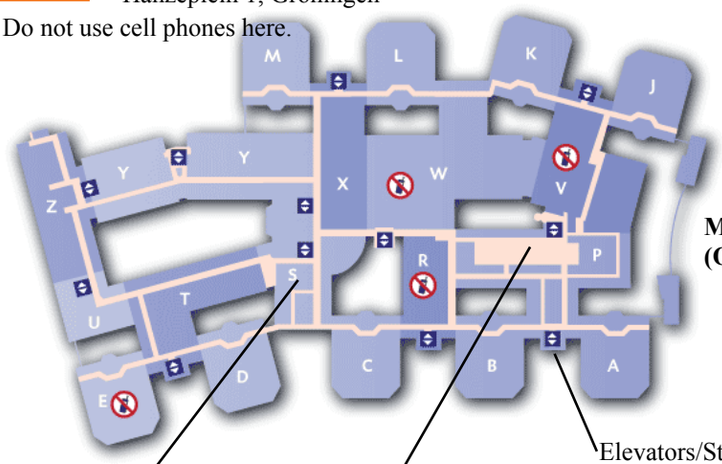


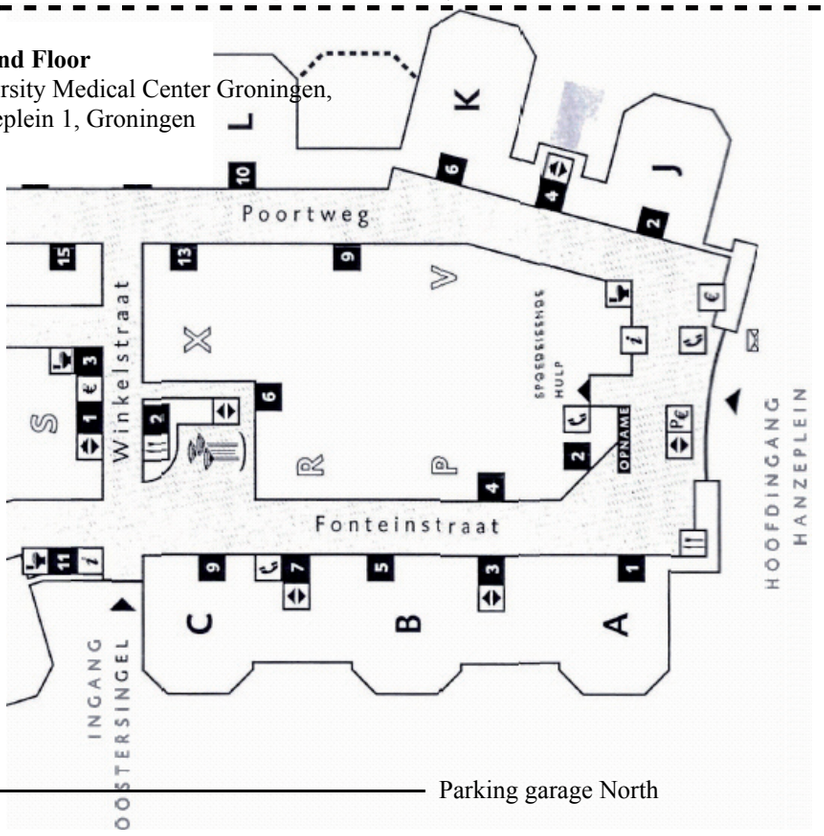
Main Entrance
(On Ground Floor)

Elevators/Stairs

Red Room /Rode Zaal
(Second floor, presentations)

Blue Patio/Blauwe patio
(First Floor, Posters and refreshments)

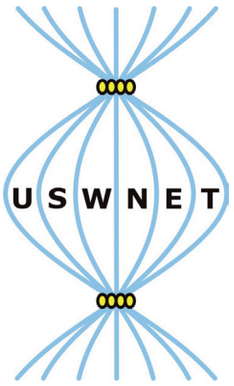
Ground Floor
University Medical Center Groningen,
Hanzeplein 1, Groningen



Main Entrance
(On Ground Floor)

Parking garage North

USWNET



ABSTRACT - Invited lecture

Saturday, October 2nd

Continuous Separation of Micro/Nano Structures Using Microfluidic Systems

Minoru Seki

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Separation of small spheres or structures based on their physical or chemical properties is one of the inevitable and crucial techniques in the fields of biological research and medical care. A number of studies have been reported concerning separation of particles, cells, droplets, micelles, vesicles, polymers or supramolecules using micro/nanofluidic devices. However, it can be said that versatile and useful microfluidic systems for sorting of micro/nano structures have not been fully established.

1. Pinched Flow Fractionation (PFF)

We have proposed a concept of 'Pinched Flow Fractionation (PFF) [1], and achieved the successful separation of various types of sub-micron to micron-size structures including polymer beads, cells, droplets and macromolecules. In this method, particles can be separated according to their diameters simply by introducing liquid flows with and without particles into a microchannel (see Fig.1), i.e., no outer field controls are needed. Also, in PFF method, separated particles can be continuously collected using multiple branch channels connected to the pinched segment.

If the multiple collection channels are uniformly arranged in the PFF method, all branch channels are not effectively used since the liquid flow in the pinched segment is equally distributed to branch channels. To overcome this disadvantage, "asymmetric pinched flow fractionation (AsPFF)" [2] was proposed. The microfluidic device for AsPFF has a short drain channel, and a large portion of the liquid flow goes through the drain channel. Liquid flow containing particles is then efficiently distributed to other branch channels. Therefore, almost all branch channels can be effectively used for particle collection.

The effluent positions of particles in PFF are determined by the microchannel structure designed beforehand and it is difficult to change the size range of the separated particles after fabrication. Therefore, we proposed "tunable pinched flow fractionation (tunable PFF)" [3], in which the effluent positions of the particles can be precisely tuned by controlling the flow rates distributed to each outlet using microvalves. And alternative method of precise tuning of PFF liquid flow using electroosmosis made a prompt change of flow distribution in the microchannel [4]. Using PFF microchannel systems, micro-droplets were also separated based on their sizes only by introducing the emulsion from the inlet of the devices [5].

2. Hydrodynamic Filtration (HDF)

We have proposed another scheme for continuous separation of particles or soft matters utilizing laminar flow profiles in a microchannel network with multiple branch channels, named Hydrodynamic Filtration (HDF) [6], in which both concentration and classification of particles can be performed at the same time, by simply introducing a liquid containing a particle mixture. In this method, the hydrodynamics of laminar flow is also utilized in a microchannel having multiple branch points and side channels, not necessitating any outer field controls. In addition, not the cross-sectional size of the microchannel, but the flow profile inside the channel determines the size limit of the concentrated/classified particles. Therefore, a problem of channel clogging, which is similar to the mesh clogging or membrane fouling in the case of existing filtration methods, can be avoided.

As shown in Fig. 2, when the relative flow rate distributed into side channels at a branching point is low enough, only a small portion of the liquid near the sidewall is withdrawn from the main stream. In this flow state, particles whose diameter is larger than a specific value cannot enter the side channels, even if they are flowing near the sidewalls in the main channel, and even if the particle is smaller than the cross sectional area of the side channels. In addition, after passing through the branching point, particles are slightly concentrated. On the other hand, when the flow rates distributed into the side channels are increased a little larger, particles near the sidewalls can go through the side channels. Particle concentration and classification can be achieved by combining these flow states.

By repeating the flow state so as that the introduced particles would never enter the side channels, a large portion of the introduced liquid can be removed from the main stream after passing through the multiple branching points, resulting in the concentrated particle alignment onto the sidewalls. Then by increasing the relative flow rates into the side channels stepwise in the downstream region, aligned particles are collected independently according to their sizes. This operation dramatically increases the collected particle concentration, since only a small portion of the liquid flow near the sidewalls is collected.

However, the inevitable contamination of small particles into the concentrated large-particle fraction decreases the separation efficiency. To overcome these disadvantages, we proposed an improved HDF scheme [7] for particle separation, employing flow splitting and recombining. That is, by repeating liquid removal from the main channel, and recombining to the downstream, particles are perfectly aligned on one sidewall in the main channel. Then, particles can be separated according to their sizes, by increasing the relative flow rates into the outlet channels. Using this modified HDF method, various kinds of microspheres or cells were classified [8], concentrated or focused [9]. Also, this scheme enables the millisecond reagent treatment against cells or particles by a simple operation [10].

Acknowledgements

I would like to thank Prof. Masumi Yamada of Chiba University. This study was supported in part by Scientific Research A (20241013) from the Ministry of Education, Culture, Sports, Science, and Technology, Japan.

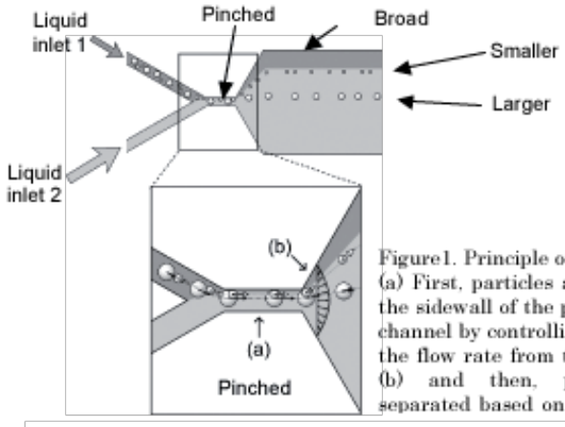

Figure1. Principle of PFF method: (a) First, particles are aligned on the sidewall of the pinched micro-channel by controlling the ratio of the flow rate from the two inlets, (b) and then, particles are separated based on their sizes in

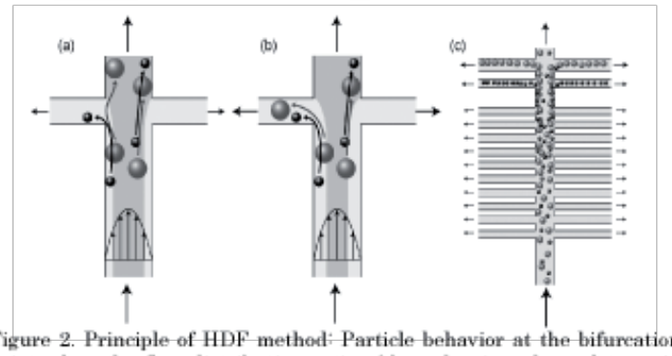

Figure 2. Principle of HDF method: Particle behavior at the bifurcation point when the flow distribution ratio of branch micro-channel to main channel is relatively (a) smaller and (b) larger. (c) Particle concentration and separation behavior in the micro-channel with multiple bifurcation

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ABSTRACTS - Session 1
Saturday, October 2nd

Sonotweezers: Towards 3D particle positioning

P. Glynne-Jones[1*], C. Demore[3], C.R.P. Courtney[2], C. Ye[1], C.-K. Ong[2], Y. Qiu[3], R.J. Boltryk[1], P.D. Wilcox[2], M. Hill[1], S. Cochran[3], B.W. Drinkwater[2].

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[2] Mechanical Engineering, University of Bristol, BS8 1TR, U.K.

[3] Institute for Medical Science & Technology, University of Dundee, DD2 1FD, UK

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Introduction

Ultrasonic particle manipulation is typically used to move particles towards locations that are fixed by the enclosure resonances and hence geometry of a device. A number of methods have been investigated for overcoming this limitation, including mechanical relative movement between transducer and chamber [1], frequency modulation [2], multiple transducers[3], and mode super-position [1, 4].

We describe in this paper some initial results representing progress in two alternative manipulation techniques.

A: Composite linear array switching

The concept, proposed by Kozuka [3], involves mounting a chamber over an array of transducers, and controlling position by selectively activating transducers. Key to the operation of this device is localising the excited resonance to the intended chamber region.

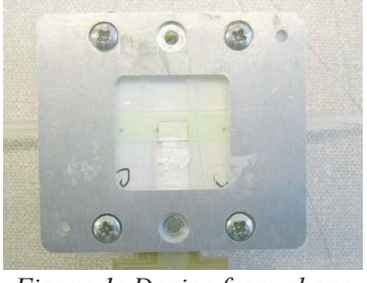

Figure 1: Device from above

A transducer array has been created by dicing a PZT/epoxy composite transducer with a wafer-dicing saw to define array elements. The 6 mm long PZT plate is diced into 30 elements with a pitch of 0.2 mm and length of 4 mm. The transducer is embedded in a glass-bubble filled epoxy substrate. A flex-circuit makes connection to the array elements. A glass capillary of inner dimensions 6 x 0.3 mm^2 with 0.3 mm thick walls is coupled to the surface of the transducer array.

A plane strain model of the device was constructed, and the resulting radiation forces calculated from the Gor'kov equations as previously described [5]. It is found that the potential energy contributions to the radiation force lead particles to a corrugated pressure node in the chamber centre. Since there is zero pressure gradient along this node, it is the (typically weaker) kinetic energy terms that lead to the lateral manipulation, as seen in Figure 2. We demonstrate experimentally the action of the device, by activating pairs of electrodes and recording the movement of polystyrene beads as the electrodes are switched.

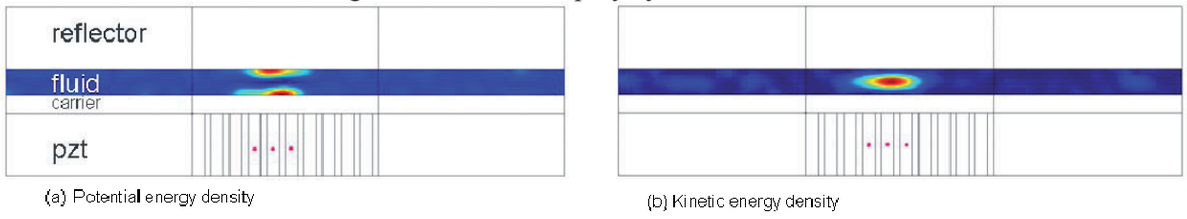

Figure 2: Modelled energy density distributions, arbitrary scale. Pink dots show which pzt elements are being driven. Dense particles experience a force towards the minimum (blue) of the potential energy density, and a weaker lateral force towards the kinetic energy density maximum (red).

Towards Integrated Acoustic Cell Trapping and Polymerase Chain Reaction

Brian L. Poe[1], Björn Hammarström[2],
Thomas Laurell[2], Johan Nilsson[2], and James P. Landers[1,3,4]

Depts. of [1]Chemistry, [3]Pathology and [4]Mechanical Engineering, University of Virginia, USA
[2]Dept. of Electrical Measurements, Div. of Nanobiotechnology, Lund University, Sweden

Introduction

Bacterial food-borne pathogens such as *Escherichia coli* pose a significant threat to consumer health, and technologies for rapid, sensitive screening of the food supply are needed to help prevent deadly outbreaks. Polymerase chain reaction (PCR) methods are extremely useful in detecting and identifying pathogens, but upstream processing and enrichment of large volume samples can limit their utility[1]. This work proposes a method of on-line cell enrichment via acoustic trapping followed by rapid non-contact PCR to detect pathogens in dilute wastewater samples. Previous studies have demonstrated efficient trapping of cells in glass microdevices[2,3], or single-step cell lysis and PCR[4]. To our knowledge, these two processes have never been combined in a single chamber, which would simplify the microchannel architecture significantly and facilitate automation.

Experimental

Two different devices were used for trapping in this work. To demonstrate acoustic trapping of *E. coli* cells, we used a 10 MHz piezoelectric transducer coupled to a rectangular glass capillary with a vertical inner dimension of 200 µm[2]. A solution of fluorescently stained bacteria was flowed through the capillary device and the transducer was actuated at several frequencies, including 10.8 MHz for a triple node trapping arrangement.

A second device, in effect a PCR-capable acoustic resonator, consisted of a 6 MHz piezoelectric transducer coupled to a microchip designed for rapid PCR. The PCR microchip was fabricated from borosilicate glass using conventional wet chemical etching[5], with dimensions as described in figure 1. During PCR, these devices have shown temperature ramp rates averaging −24.9 ºC/s for cooling and 17.1 ºC/s for heating[5] – values that are an order of magnitude faster than conventional thermocyclers. The PCR chambers were simulated using the acoustics module COMSOL multiphysics application as shown in fig. 1c.

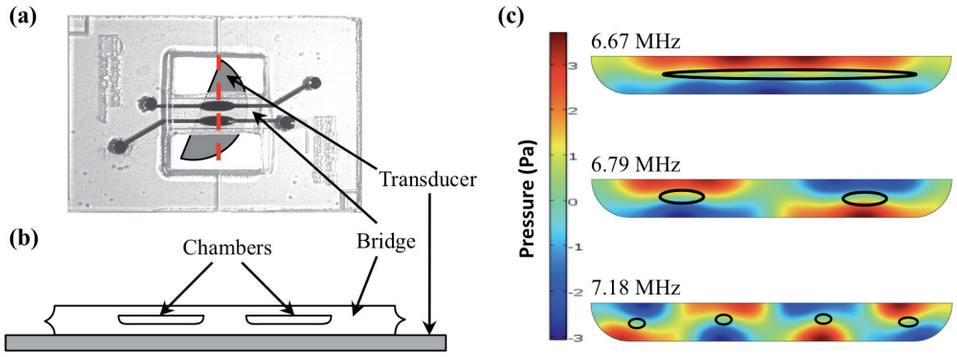

Fig. 1. *(a) Photograph of the bridge PCR chip used in this work. Dimensions: 20 mm x 30 mm. Bridge region is 4 mm wide, and chambers are 1 mm wide by 3.175 mm long and 113 µm deep with ~115 µm of glass above and below the chambers. (b) Cross-sectional view of the bridge region (along dotted line in panel a) that shows placement of the piezoelectric transducer. (c) A single chamber was modeled using COMSOL multiphysics simulation software. Potential low pressure trapping nodes are indicated with black ovals for various modes.*

Frequency-shift-keying actuation with a damped transducer for ultrasonic particle aggregation in a multi-well chip

Mathias Ohlin, Thomas Frisk, B. Önfelt and Martin Wiklund

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Introduction

Positioning and aggregation of bioparticles are important for cell-cell interaction studies. We have previously demonstrated parallelized aggregation of immune cells in a multi-well chip having quadratic (300×300 µm) wells, by the use of a frequency-modulation actuation technique [1]. In that work, we performed linear frequency sweeps within a relatively narrow band (<100 kHz), in order to obtain a uniform and predictable trapping effect in all 100 wells. In this abstract, we investigate the use of frequency-shift-keying actuation [2] by mixing the two lowest resonance frequencies of rectangular (350×460 µm) wells. We also investigate a novel type of damped wedge transducer for particle manipulation purposes compatible with wide-band (>100 kHz) frequency modulation.

Experiment

The transducer, see Fig. 1a-b, consists of a titanium wedge, PZT piezo crystal, and an epoxy backing layer. The multi-well chip is fabricated out of silicon and glass in which 100 rectangular wells have been etched. In the experiment the wells were filled with a 1 % TWEEN-20 water suspension of 10 µm fluorescent polystyrene beads.

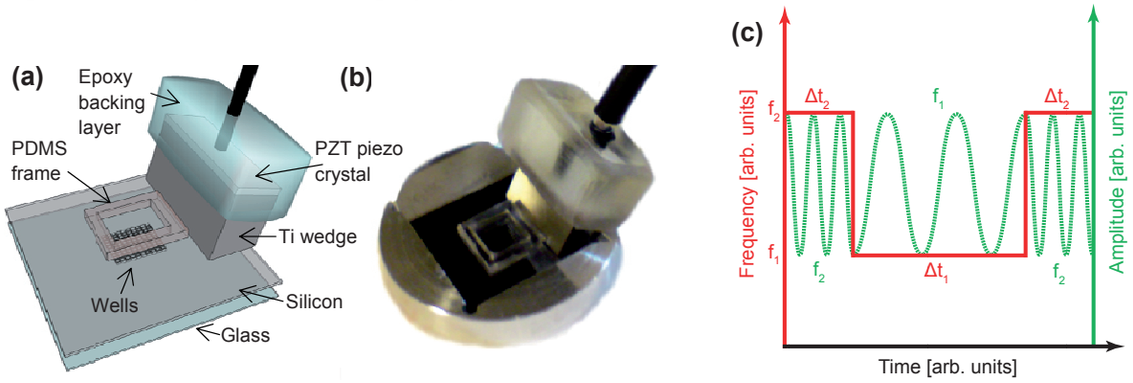

Fig 1. (a) The silicon-glass multi-well chip. (b) A photograph showing the experimental setup. (c) Diagram showing the principle of the frequency-shift-keying actuation method used.

ABSTRACTS - Session 2
Saturday, October 2nd

Quantifying acoustic streaming in large-particle acoustophoresis

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[b]Dept. Measurement Technology & Industrial Electrical Engineering, Lund University, SWEDEN

We present measurements of the small acoustic streaming component in large-particle acoustophoresis dominated by the acoustic radiation force. The experiments are carried out by using a novel micrometer-resolution particle-image velocimetry (micro-PIV) setup capable of automated full-chip measurements of microchannel acoustophoresis. The system allows for fast collection of a sufficient amount of data enabling for the first time reliable determination of the forces from acoustic radiation and acoustic streaming acting simultaneously on microbead suspensions in a microfluidic system. Our work is an important step in the further development of using the combination of ultrasound and microfluidics [1–5] to manipulate nm-sized biomolecules.

A particular challenge, and the motivation for our work, is the cross-over of the acoustic response of a particle/cell with radius a from the well-understood radiation force $\boldsymbol{F}_{\mathrm{ac}}$ to the ill-characterized acoustic streaming force $\boldsymbol{F}_{\mathrm{str}}$. In the standing one-dimensional plane wave assumption $u(y) = u_0 \sin(ky)$ in a liquid of viscosity η, density ρ, and speed of sound c, the force amplitudes read $|F_{\mathrm{ac}}| = \pi\rho k a^3 \Phi u_0^2$ and $|F_{\mathrm{str}}| = 6\pi\eta a \Psi u_0^2/c$, where Φ is the material dependent contrast factor, whereas Ψ is a geometry dependent streaming factor. This leads to the acoustic radiation force dominating for $a \gtrsim 1$ µm and the acoustic streaming force dominating for $a \lesssim 1$ µm [6, 7]. Moreover, the induced particle/cell velocity due to $\boldsymbol{F}_{\mathrm{ac}}$ and $\boldsymbol{F}_{\mathrm{str}}$ scales as a^2 and a, respectively.

We use our micro-PIV setup to accurately record the acoustophoretic velocity field $\boldsymbol{u}(x, y)$ for suspensions of 5, 3, and 2 µm polystyrene microbeads in concentrations of 0.05, 0.025, and 0.0125 % w/v, respectively. We use the same silicon/glass chip and acoustic 2-MHz-actuation under stop-flow conditions as in Ref. [8]. Theoretically, the acoustic radiation force $\boldsymbol{F}_{\mathrm{ac}}$ is a gradient force, while the acoustic streaming force $\boldsymbol{F}_{\mathrm{str}}$ is a rotation force due to viscous drag from the vortex motion $\boldsymbol{u}_{\mathrm{str}}$ in the carrier liquid. From force balance with the Stokes drag force, we obtain the microbead velocity $\boldsymbol{u} = \boldsymbol{F}_{\mathrm{ac}}/(6\pi\eta a) + \boldsymbol{u}_{\mathrm{str}}$, i.e., a sum of a gradient field and a rotation field. Employing discrete Fourier transformation and Helmholtz decomposition we are able to decompose the measured velocity field $\boldsymbol{u} = \boldsymbol{u}_\phi + \boldsymbol{u}_\psi = \boldsymbol{\nabla}\phi + \boldsymbol{\nabla} \times [\psi \boldsymbol{e}_z]$, where the gradient term is rotation-free and the rotation term is divergence-free. The decomposition allows us to *in situ* measure the potentials ϕ and ψ for the acoustic radiation and acoustic streaming, respectively. We then determine the weight of the two velocity components in terms of the 2-norm $||\boldsymbol{u}||^2 = \int \mathrm{d}a\, \boldsymbol{u} \cdot \boldsymbol{u}$. In Fig. 1 is shown the extracted potentials for the focusing of the 3 µm beads. In Fig. 2, the velocity fields and magnitudes of $\boldsymbol{u}_\phi$ and $\boldsymbol{u}_\psi$ are shown for all three bead samples, such that the colorplots scale as the expected a^2 and a.

In Table 1 is shown that as the microbead diameter decreases from 5 to 2 µm the weight of the acoustic radiation decreases from 0.98 to 0.92, while the weight of the acoustic streaming increases from 0.11 to 0.34 . Moreover, the expected size scaling of $\boldsymbol{u}_\phi$ and $\boldsymbol{u}_\psi$ is observed, but the results from the focusing of the 2 µm beads deviates slightly as diffusion plays a larger role for smaller particles. The numerical error from the decomposition of the velocity field is around 4% and is thus a minor uncertainty to the obtained results.

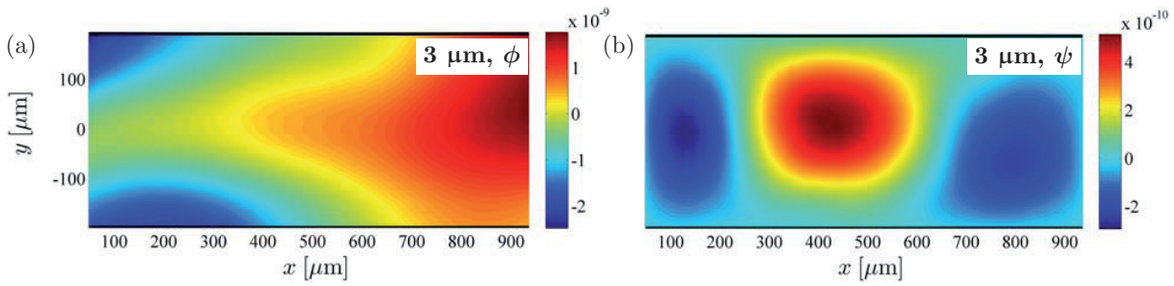

Figure 1: Color plot of the acoustic potentials extracted from acoustophoresis on 3 µm beads. (a) The scalar potential $\phi(x, y) \approx 2 \times 10^{-9}$ m^2/s. (b) The vector potential $\psi(x, y) \approx 3 \times 10^{-10}$ m^2/s.

Circular flow of liquid induced by acoustic streaming

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Introduction

For biomedical applications there is interest in contactless handling of fluid volume within a closed compartment to avoid contamination of the sample. A net mean fluid flow can be generated with acoustic streaming[1], fulfilling these requirements. The streaming can be used for mixing or for ultrasonic pumps[2] or to create a circular flow with or without mixing. A device with a circular flow allows for transport of e.g. particles suspended in the fluid and can be combined with a particle trap based on e.g. acoustic radiation forces. Such a device has been investigated experimentally as well as with a numerical simulation.

Method and setup

The plastic chamber, shown in Fig. 1 and Fig. 2, has the outer dimensions of 40 x 20 x 3 mm and consists of a 1.5 mm PMMA frame covered by two 0.1 mm polyester foils. The device is filled with water and excited with a 10 x 3 x 2 mm piezoelectric transducer on the lower left wall. The lower right corner is replaced by a tilted wall to increase the path length of the emitted plane wave by deflection. This is done, because the device is designed to have as a main driving mechanism plane travelling wave streaming.

Numerical simulation

The device is simulated in 2D with a fluid-solid interaction model shown in Fig. 1.

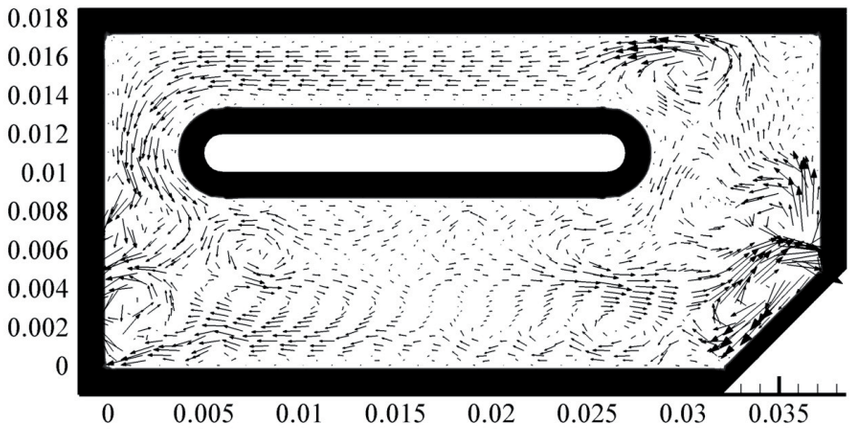

Fig. 1: Fluid velocity field (293 kHz)

The chamber frame is regarded as elastic solid with a thickness of 1.5 mm enclosing the inner fluid. The linear elastic dynamic equations for the chamber and full Navier-Stokes equations for the fluid

are coupled with the interactive boundary conditions of velocity and stress between the solid and fluid and solved simultaneously in the calculations. The excitation is introduced with a velocity boundary condition. The numerical algorithm is based on the finite volume method (FVM) with triangular meshes described in our previous paper[3]. The time-averaged streaming can be calculated by averaging the fluid velocities computed from the full Navier-Stokes equation over several cycles. Figure 1 shows the acoustic streaming pattern at 293 kHz, with a laminar flow in the negative x-direction in the region above the chamber hole and several vortices close to the excitation and the deflection wall (right corner).

Experiments

Driving the piezo at its resonance (293 kHz), strong streaming has been observed. Some streamlines indicating the flow field are shown in Fig. 2. In the upper region, the laminar flow character, the direction and the order of magnitude of the highest fluid velocity correlates with the numerical simulation, while in the lower region both experiment and simulation show vortices. The highest fluid velocities in the laminar region are in the order of 5 mm/s. In addition to the streaming patterns in the lower region a standing wave is established in the y-direction forcing particles into parallel lines by acoustic radiation forces.

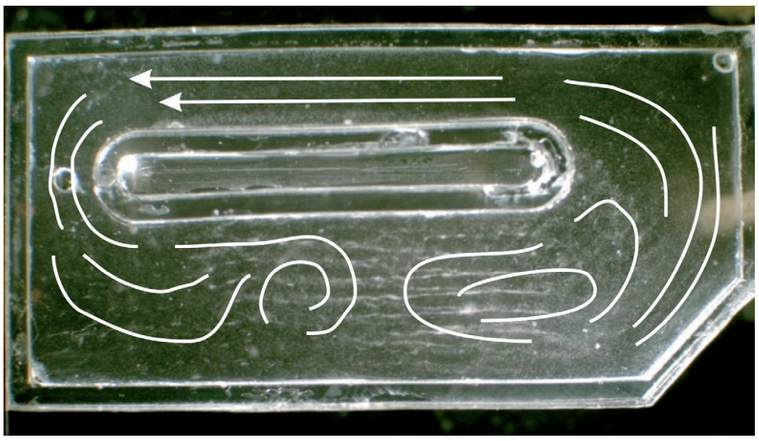

Fig. 2: Streamlines (293 kHz)

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Reduction of the influence of the sample fluid density over the resonance frequency of the acoustic modes in a microfluidic cavity

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In recent years there has been increase in the microfluidic manipulation of micro and nanoparticles and biological samples by use of acoustical interactions. Compared to the traveling wave modes, standing acoustic waves are dominantly used due to the fact that acoustic radiation forces are significantly higher in the case of standing wave modes [1]. Since acoustical interactions do not heavily depend on pH or electrolyte concentration, applicability in wide range of samples -independent of their electrokinetic properties- is possible. However, one major limitation is that as the fluid sample filling the cavity serves as the medium for the wave propagation, as shown in Figure 1, it is one of the major parameters influencing the resonance frequency, f_n, through the relation; $f_n = \frac{n}{2l}\sqrt{\frac{K}{\rho}}$, where n is the harmonic number, w is the width of the cavity, K is the bulk modulus of the fluid, and ρ is the density of the fluid. This factor limits the device performance in a way that, the resonance frequency of the microfluidic actuator is needed to be calibrated with respect to the density and the bulk modulus of the fluid sample. In addition, variations in the number of particles suspended in the fluid makes the problem more sophisticated.

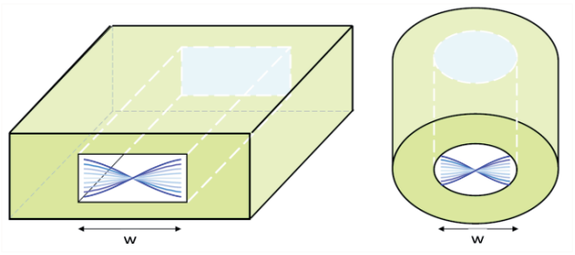

Figure 1. *Resonance frequency of the acoustic standing wave modes generated in a microfluidic cavity heavily depends on the cavity dimensions and the bulk modulus and the density of the material filling the cavity.*

We had previously showed that by using flexural vibrations on a polyimide coated standard silica capillary (Figure 2), stationary acoustic modes can be generated. This actuation mechanism leads to micro and nanoparticle focusing and separation at low power drives. Harmonicity of these coupled acoustic modes is governed by a more sophisticated formula;

$$f_n = \frac{87}{100}\frac{\frac{(\beta_n l)^2}{2\pi l^2}\sqrt{\frac{E_1 I_1 + E_3 I_3}{\rho_1 A_1 + \rho_2 A_2 + \rho_3 A_3}}}{\sqrt{1 + \frac{(\beta_n l)^2 k^2}{l^2}\left(1 + \frac{E}{G\kappa}\right)}}$$

where l is the length of capillary, E is the Young's modulus, I is the area moment inertia, k is radius of gyration, G is the shear modulus, κ is a constant related to the shear properties of the capillary, ρ is the density, A is the cross-sectional area, and among the subindices, 1 represents the silica, 2 represents the fluid medium and 3 represents the polyimide coating. In

ABSTRACTS - Session 3
Saturday, October 2nd

Acoustophoretic contrast factors of living cells measured in acoustic fields calibrated *in situ* using polystyrene microbeads

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We report a new method, which allows for accurate measurement of the acostophoretic contrast factor Φ of different cell types, an acousto-physical parameter of fundamental importance in microchip acoustophoresis. This progress has been powered by the recent breakthrough in measuring the absolute acoustic energy density in microchannels presented by Barnkob *et al*. [1] and is a long sought advancement since acoustophoretic separation of biological cells is utilized in a growing number of applications [2-7]. Prior to this work there has been no systematic measurements of Φ appearing in the expression for the acoustophoretic force $F_{ac} = \Phi\, k_y\, V\, E_{ac} \sin(2k_y y)$, where k_y is the transverse wavenumber, V the cell volume, E_{ac} the acoustic energy density, and y the transverse position.

We utilize a silicon/glass chip from Ref. [1] containing a straight microchannel of length 40 mm, width 377 μm, and height 157 μm. Upon actuation at a frequency f = 1.970 MHz and piezo voltage U_{pp} = 1.0 V, the microchannel supports a strong 1D transverse half-wavelength acoustic resonance. The acoustophoretic force F_{ac} focuses the particles/cells in the solution to the vertical center plane of the channel.

The channel is initially filled with a solution of 5 μm polystyrene microbeads (PS) of known acoustophoretic contrast factor Φ_{PS}. To determine F_{ac} *in situ* we first use a full-image micron-resolution particle image velocimetry (micro-PIV) setup to measure the acoustophoretic 2D velocity field $\boldsymbol{u}(x, y)$ of the PS microbeads, see Fig. 1(a), which shows that in the field-of-view, the velocity field is close to the ideal 1D transverse resonance form $u_y = u_0 \sin(2k_y y)$. We then insert the average u_y of the y-component in the force balance $F_{ac} = 6\pi\eta a u_y$ to determine the magnitude of the acoustic energy density E_{ac} and the transverse wavelength λ_y as in Ref. [1]. Subsequently we fill the channel with cells and measure their trajectories $y(t)$, see Fig. 1(b), which shows an image of the MESC-diff4d cells undergoing the acoustophoretic motion towards the center of the channel.

To extract the contrast factors Φ we first measure each cell radius a optically, see Fig. 2(a), fit each trajectory $y(t)$ to determine individual factors $a^2\,\Phi$, see Fig. 2(b), and finally extract the acoustophoretic contrast factors Φ, see Fig. 2(c). Our first biological experiments were performed on a human embryonic ventral mesencephalic cell line (MESC2.10) [8]. When cells are differentiated in special medium for 4 days (MESC-diff4d), the cells will adopt a mature neuronal phenotype with robust dopaminergic characteristics (eg. express tyrosine hydroxylase and dopamine transporter). The goal was to find out if undifferentiated and differentiated cells display differences in acoustic contrast factor that will allow for acoustophoretic separation. In Table 1 all measured parameter values for the two cell type experiments are summarized.

We have introduced a new method for accurate measurement of acoustophoretic contrast factor Φ. The method has been tested on two cell populations and demonstrated that they despite a close acoustic similarity are nevertheless separable due to different Φ-distributions.

Ultrasonic extraction of tumour cells from peripheral blood in a plastic resonator

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This paper reports for the first time a new method based on the application of ultrasounds for isolating circulating tumour cells (CTC) as a non-invasive tool for the detection of recurrent disease, for monitoring and therapy selection.

The use of parallel flows in combination with the ultrasound actuation allows selective separation and extraction of the tumour cells exploiting the difference of size/density. This technique is based on the action of a radiation force and does not require the introduction of any external element in the sample, allowing the recovering of intact cells for later bio-molecular analysis.

The paper reports also an improvement of a technology recently developed [1] and tested with inorganic suspensions IEEE '09 [2]. A new configuration for the chip includes an air shorted chamber parallel to the channel of treatment to provide a more precise control of the location where cells collect. The device differs from the relevant ultrasonic separators of the literature. Unlike them [3-5], based on the resonance of the channel, it behaves as a multilayer system in which all the layers are involved in the establishment of a standing wave across its whole width. An schematic view of the device and its work is shown in Figure 1.The strategy presented in this device allows the location of the particle collector at a desired position within the channel, varying slightly the frequency around 1MHz. The device was fabricated by standard SU-8 photolithography using PMMA as substrate.

For the sample preparation, known numbers of tumour cells from PANC-1 cell line of pancreatic origin were added to 5 ml of peripheral blood from a healthy donor to be treated by ultrasounds within the channel of the microdevice. Red blood cells were lysated using Buffer EL (Erythrocyte Lysis Buffer, Qiagen) in order to reduce the blood viscosity, minimizing its behaviour as non-newtonian fluid. The blood-sample containing tumour cells flows in parallel with PBS 10%-diluted in distilled water along the channel during the acoustic treatment.

This paper presents results of selective separation of tumour cells under the ultrasounds. At a frequency of 1.023MHz a node of pressure is established inside the channel close to the left sidewall, where larger cells collect, as shown in the photographs of Figure 2.

In the photograph of Figure 3 CTCs, leave the channel through the lower outlet, leaving the device separated from the blood sample. The cell behaviour were filmed with a high-speed CCD camera at capture speed of 250frames/second resolving the single cells at a flow a rate Q=20μL/min. This article reports for the first time in the literature the feasibility of ultrasounds to isolate circulating tumour cells from peripheral blood. It is the first step in a feasibility-study

Numerical modelling of a quarter wave separator for the removal of lipid particles from pericardial suction blood

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Introduction

The removal of lipid particles from pericardial suction blood (PSB) collected during cardiac surgery before re-transfusion to the patient is essential in preventing post-operative cognitive disorders related to the build-up of lipid particles in the brain [1], [2]. An effective separation technique that allows the removal of lipid particles from blood over a range of sizes is needed. Previous work investigated the use of ultrasonic standing waves (USW) in microstructured flow channels to achieve the separation [3]. This work introduces the use of a novel type of ultrasonic separator, a quarter wave radial inward flow separator. This separator can be scaled-up by just increasing its diameter. This presents a great advantage in respect to operating with several channels in parallel, which can lead to a lack of homogeneity of processing conditions between the different channels. A CFD (Computational Fluid Dynamics)-based model taking into account both the flow configuration in the separator and the forces acting on the particles was developed and used as a design tool for the radial flow separator.

Separator configuration and model

The radial flow separator considered in this work is shown schematically in Figure 1. Blood flows in the gap between two discs from a peripheral inlet towards the centre outlet while the quarter wave system is established along the main axis of the separator, in the gap between the discs. The quarter wave configuration allows the separation of the lipid particles on the matching layer side of the separator and of the red blood cells (RBCs) on the reflector side of the separator. As the flow exits the separator on the reflector side, this favours the recovery of RBCs with the purified blood.

The computational platform CFD-ACE+ (ESI Group, Paris) was used to model the flow characteristics, in the absence of particles at first. The purpose of this initial investigation was to ensure that recirculation areas and the ensuing re-mixing is safely eliminated, leading to the prevalence of streamlined flow in the separator.

ABSTRACT - Invited lecture

Microscale electrical manipulation of cells

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Cells are the fundamental unit of life, and as such, understanding how they function and exploiting their features has garnered intense interest. From basic biologists trying to understand how cells work to clinicians trying to detect the presence of tumor cells during cancer therapy, many communities have intense interest in manipulating, measuring, and separating cells. Among methods to separate cells, intrinsic separation techniques, which rely on intrinsic properties of cells to conduct the separation (e.g., density, opacity, electrical properties) are attractive as they can be simple to implement and require minimal pre-processing of the cells. However, these techniques are only useful if the biological property of interest (e.g., metastasis) correlates with the physical property chosen for the separation. Understanding the correlation between intrinsic phenotypes and biological phenotypes is thus important in exploiting the power of intrinsic separation technologies.

Our lab uses microscale electrical methods to enact separations of cells. Cells can be modeled electrically as a series of RC-networks embodying the properties of the cell interior (the cytoplasm), the cell membrane, and the exterior of the cell (the media). If any of these electrical parameters varies in response to a biological difference, then the overall electrical properties of the cell will vary, which can be detected electrically. Many important biological differences are already known to manifest themselves as electrical differences. For instance, as cells die via apoptosis, the membrane properties change (both conductance and capacitance), which can be readily detected electrically [1].

To detect electrical properties of cells, we have developed a method called iso-dielectric separation (IDS), where cells are placed in a spatially varying electric field and a spatially varying conductivity gradient that forces them to the location in a channel where their net polarizability (p) is zero [2]. At that location, the force on the cell, given by the dielectrophoretic force ($F = p \cdot \nabla E$) will go to zero [3, 4]. Cells with different polarizabilities will end up in different locations in the channel, thus separating cells based upon electrical properties. The physical implementation of the method consists of a glass chip with patterned electrodes to create the field, along with microfluidics to deliver cells and create the conductivity gradient. We have used the device to separate and measure the electric properties of bacteria, yeast, and mammalian cells.

Iso-dielectric separation has a number of features that exploit the properties of microsystems for cell manipulation. First, the conductivity gradients in the system can interact with the imposed electric fields to create electrohydrodynamic instabilities. These instabilities are stabilized by the walls, so scaling down the system via microfabrication enhances performance. Second, the system is capable of manipulating ~μm-sized cells one-at-a-time at high throughput (~10^4 cells/min), enabling both single-cell resolution yet the ability to process significant numbers of cells. Third, the method is fundamentally insensitive to non-electrical variations in the cells. The most important such variation is size variation. A typical cell population will have an ~2× variation in size, and so avoiding spurious size-based separations is paramount. Cells in IDS migrate to their electrical equilibrium point, which is independent of cell size, and thus the separation is independent of size. Finally, because the separation is electrically based, the ability to integrate this approach with more sophisticated electronics for sensing/actuation is readily possible.

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Theoretical and numerical calculations for the time-averaged acoustic forces and torques experienced by a rigid elliptic cylinder in an ideal fluid

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Introduction

Particle manipulators using ultrasound standing waves have developed significantly in recent years. The particles are not only able to be positioned and separated, but also rotated by these novel devices. T. Schwarz et. al.[1] reported an elaborated device to rotate a non-spherical particle by amplitude modulation of excitation voltages. It is necessary to accurately predict the acoustic radiation force and torque acting on the particles in a sound field to design and improve ultrasonic particle manipulators. Former approaches mainly concentrated on spherical and cylindrical particles and also Gor'kov's theory is not suitable for particles of large aspect ratio. In this paper, we present a theoretical calculation for acoustic radiation force and torque on an elliptic cross section in an ideal fluid with a sound field as well as a numerical simulation based on the numerical scheme proposed in our previous paper [2].

Theory

The incident and scattered velocity potentials may be expanded with Mathieu's functions ce_n and se_n as well as radial Mathieu's functions Mc_n and Ms_n [3] in the elliptical coordinates (ξ,η) whose origin locates at the elliptic cross section center,

$$\varphi_I = \sum_{n=0}^{+\infty} a_n Mc_n^{(1)}(\xi,q)ce_n(\eta,q) + \sum_{n=1}^{+\infty} b_n Ms_n^{(1)}(\xi,q)se_n(\eta,q), \tag{1}$$

$$\varphi_S = \sum_{n=0}^{+\infty} c_n Mc_n^{(1)}(\xi,q)ce_n(\eta,q) + \sum_{n=1}^{+\infty} d_n Ms_n^{(1)}(\xi,q)se_n(\eta,q), \tag{2}$$

where coefficients a_n and b_n are obtained from the incident wave field and coefficients c_n and d_n are determined by the boundary conditions on the cylinder surface. The total velocity potential is the sum of the incident and scattered fields

$$\varphi_T = \varphi_I + \varphi_S \tag{3}$$

The velocity components on the cylinder surface are

$$u_\xi = \frac{\partial \varphi_T}{cJ(\xi,\eta)\partial\xi}, \tag{4}$$

$$u_\eta = \frac{\partial \varphi_T}{cJ(\xi,\eta)\partial\eta}. \tag{5}$$

where c is the half distance of the two foci and $J(\xi,\eta) = \sqrt{(\cosh 2\xi - \cos 2\eta)/2}$. The time-averaged acoustic radiation force and torque are [4]

$$\mathbf{F} = -\frac{1}{2}\int_{S_0}\left[\frac{1}{2}\left(\frac{\rho_0}{c_0^2}\frac{\partial\varphi_T}{\partial t}\frac{\partial\varphi_T^*}{\partial t} - \rho_0(\nabla\varphi_T)(\nabla\varphi_T^*)\right)\hat{\boldsymbol{\xi}} + \rho_0(u_\xi\hat{\boldsymbol{\xi}} + u_\eta\hat{\boldsymbol{\eta}})u_\xi^*\right]\mathrm{d}S, \tag{6}$$

Hexagonal chamber for rotation of non-spherical particles

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Introduction:
The rotational manipulation of micro-particles in micro-fluidic devices is another step to expand the possible applications of ultrasonic manipulation. The rotation of non-spherical particles with amplitude modulation [1] has been shown previously, where the superposition of orthogonal ultrasonic standing waves in a square chamber was used. We want to present another method for rotation using one-dimensional ultrasonic standing waves in a hexagonal chamber.

Method:
Non-spherical particles experience a torque in an ultrasonic standing wave. Fibers shorter than one-fourth of the wavelength are constrained at the pressure node and are oriented perpendicular to the direction of wave propagation [2,3]. It is possible to use this acoustic radiation torque for a controlled rotation of objects.
The working principle is based on the alternating generation of ultrasonic standing waves with difference in the propagation direction. The hexagonal configuration of the chamber allows setting up standing waves with a wave vector oriented along three different directions in one plane. An alternating excitation of three actuators allows the complete rotation of an object in 60° steps. Figure 1 shows schematically the three force potentials created by means of each piezoelectric actuator and a fiber with its orientation in this potential.

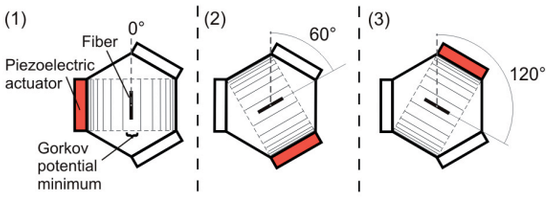

Fig.1: Actuation principle of hexagonal chamber

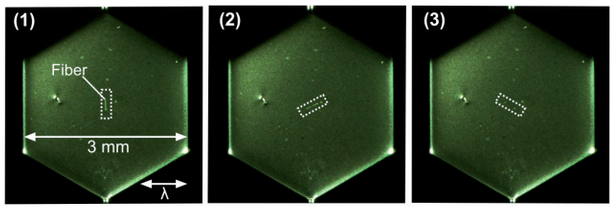

Fig.2: Experiment with glass fiber (200 µm, Ø = 15 µm)

Results:
The device consists of a hexagonal chamber with a diameter of 3 mm etched into silicon and covered with a glass plate. The actuation of the system is achieved with three piezoelectric transducers fixed at the backside of the device. The excitation frequency for all transducers was 1730 kHz. A glass fiber (200 µm, Ø = 15 µm) suspended in di-water was used as a non-spherical particle. The different orientations due to the three piezoelectric transducers can be seen in Figure 2. Clockwise and counterclockwise rotations at different angular velocities have been achieved. The maximal speed of rotation for this setup was determined to be approximately 30 rpm.

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Exploiting an USW for attenuated total reflection spectroscopy

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Introduction

The growing use of biotechnology as a manufacturing route for e.g. antibiotics and other medical compounds stimulates the development of reliable sensors for bioprocess purposes. Mid-infrared spectroscopy is increasingly popular in process analytical chemistry because of its ability to directly provide molecular specific (bio-)chemical information about a given sample. The ATR (Attenuated Total Reflection) spectroscopy is a method especially helpful in the presence of highly absorbing substances like water in aqueous solutions, only samples present within the micrometer thin evanescent field in the proximity of the ATR element contribute to the spectrum. An USW has recently been successfully exploited in a flow cell when applying the stopped flow principle – suspended particles are allowed to settle onto the horizontal ATR by stopping the flow of the suspension - to prevent the contamination of the surface sensitive optical element [1].

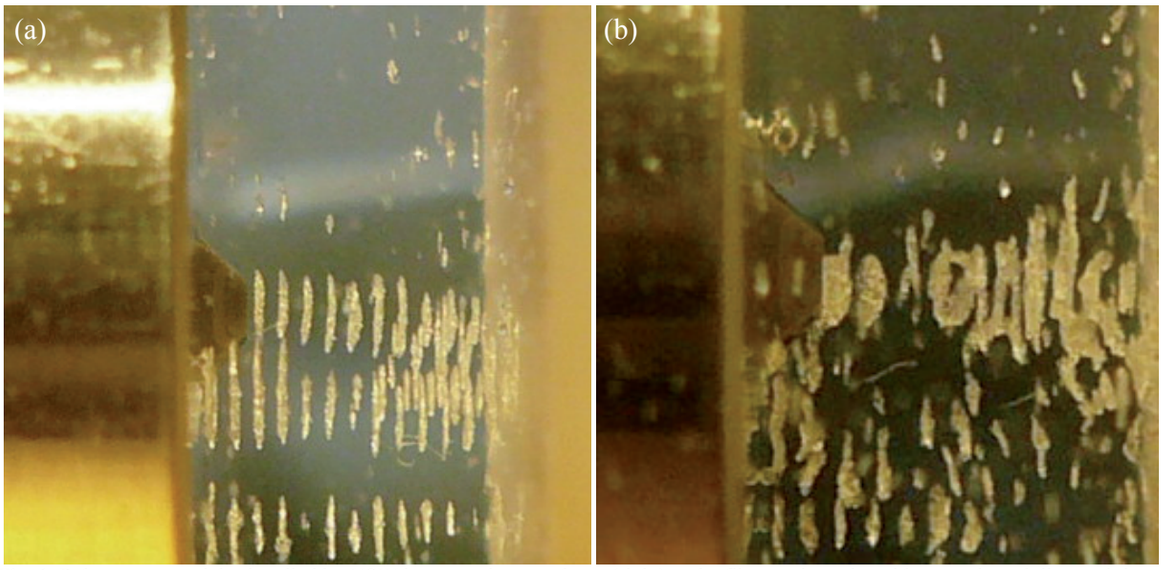

Fig. 1. Polystyrene particles handled by USW between the transducer (right) and a fibre-optic probe with cone-shaped diamond tip. (a) Particles kept from the top, (b) particles pushed

In this paper the local selectivity is utilised for sensing purposes by the application of acoustic radiation forces exerted on particles within an USW. The frequency of the ultrasonic field is used to manipulate the whereabouts of the solid fraction of a suspension relative to an ATR probe [2]. More precisely the field is used to either keep suspended particles away from the optical sensor like in Fig.1(a) or to push them towards its sensitive surface like shown in Fig. 1(b).

Acoustophoretic Preparation of Blood Components from Apheresis Product

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Introduction

The content of apharesis product is blood cells (WBCs), platelets (PLTs) and a small amount of red blood cells (RBCs). Although still a complex biolfluid, the apharesis product is a starting point for further blood component fractionation in clinical practice. Platelet preparations are a product of continuous need in transfusion medicine as PLTs can not be stored for extended periods of time (max. 5-7 days). To avoid risk of immunologically induced reactions and potentially viral infection contaminations, minimising the WBC content in the platelet fraction of the apharesis product is desired. WBC's reduced PLT preparations can be prepared by acoustophoretic processing of the apharesis product. At the same time a highly purified fraction of WBCs is obtained which can be further used for preparation of peripheral blood stem cell populations for clinical therapy. The current study investigates the possibility of employing acoustophoresis to standard clinical apharesis products to extract both purified platelet fractions as well as WBC fractions.

By utilizing the size dependency of the acoustic radiation force we have shown that it is possible to tune a resonant system such that larger cells can be fractionated from smaller.[1]

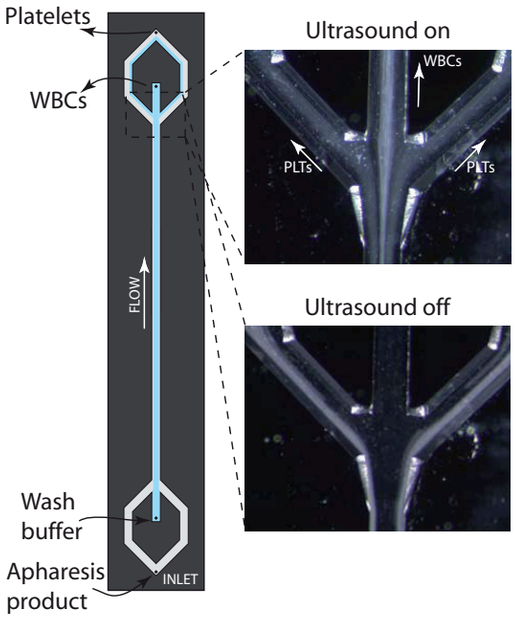

Figure 1. Schematic of the acoustophoresis chip and micrographs of the separation. The acoustic force is tuned such that only the larger WBCs are focused in the pressure node in the center of the channel. The smaller platelets (greyish mist at the side in the micrograph above) are less affected by the radiation force and their spatial position allows them to exit through the side branches.

Development of a system for label-free somatic cell enumeration in raw milk using acoustophoresis

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Denmark

Introduction

A system for acoustophoretic somatic cell enumeration in raw milk, eliminating the need for labelling and lipid solvents used in conventional milk analysis, was developed. The authors hope this will facilitate a route to miniaturized systems with integrated on-chip impedance measurement capabilities for cell enumeration. Our previous work has shown the benefits of acoustophoretic sample pretreatment for raw milk analysis [1]. Here we present proof-of-concept data to show that acoustophoresis and Coulter Counting can be combined to measure somatic cell count (SCC) levels, an important udder health indicator used in the dairy industry when analysing milk to determine live stock health and money paid to producers [2]. The use of acoustophoresis for successful separation of complex bio-suspensions has previously been shown [3]. Several groups have also reported successful culturing and high viability of cells during and after exposure to acoustic forces comparable to those used here [4,5]. We present successful acoustophoretic sorting of raw milk samples with high SCC levels, normally found in dairy cattle suffering from mastitis. Original raw milk samples were analyzed by standard flow cytometry and compared to corresponding Coulter Counter data with and without acoustophoretic processing for milk lipid removal.

Experimental

An acoustic chip was fabricated in accordance with previously reported techniques [1]. A schematic view of a complete acoustic setup along with an enlarged chip and an acoustic profiles in can be seen in Fig 1. Samples with 5.758*10^6 cells/ml entered the chip through the center inlet (35µl/min) and were hydrodynamically laminated using mQ-water (420 µl/min). Lipid depleted samples were extracted through the center outlet (60 µl/min) while lipids exited through the side outlets (390 µl/min). The small pressure surplus decrease air bubble formation. The multiple node acoustic approach (the channel is 1125 µm wide, corresponding to 3/2 λ when operated at 2 MHz) prevents lipid content from reaching channel walls which may otherwise cause clogging and erroneous operation [1]. Samples were run through the chip with and without acoustic power activated and collected in 100 µl loops during five minutes.

Results/Conclusion

Lipid depletion of the sample using acostophoresis reveals the cell peak (7-10 µm) in the Coulter Counter data, normally not visible in the raw milk sample, Fig 2. Coulter counter data for the particle sizes normally associated with cells indicated 6.3*10^6 particles/ml. This value was consistently ~0.5*10^6 higher than cytometry data of the somatic cells in the same samples (n=10). This might be due to some ≥5 µm lipid vesicle residue in the sample. The obtained data however clearly demonstrates that acoustophoresis in combination with Coulter Counter analysis offers a label- and lipid solvent free method for somatic cell counting, eliminating the need for chemical sample pretreatment. The authors hope to integrate Coulter Counter type impedance analysis on-chip in the near future.

Piezoelectric ceramics or composites for micromanipulation in multilayer plastic resonators

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Transmission of 3D-vibration modes of piezoelectric ceramics working at their thickness mode to a plastic multilayer micro-resonator designed for particle manipulation [1] has been experimentally analysed through the node of pressure established inside the channel of treatment. An ultrasonic actuator is attached to one of the external lateral edges of the plastic chip. A strategic combination of the width of the different layers of the device allows the establishment of a node of pressure inside the channel at a frequency close to that of the thickness mode of the ultrasonic actuator. The width of the channel is somewhat larger than a quarter of wavelength to collect particles with positive acoustic contrast factors ($\Phi>0$) at the node of pressure established at a location closer to one of the sidewalls (approximately one third of its width), favouring their extraction from their host suspension through a different outlet. Two types of ultrasonic actuators were attached to the chip to make the tests.

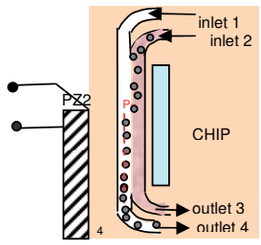

Figure 1 Scheme of the device

A PZ26 element showed a complex flexural vibration mode overlapped at its thickness resonance mode, at 1020kHz (figure 2), whose amplitude was measured by a vibrometer.

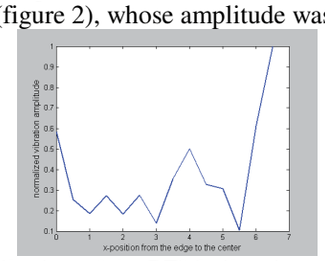

Figure 2: spatial distribution of the PZ26 vibration amplitude at f=1020kHz

On the Simulation of Acoustic Standing Waves for Microfluidic Applications with the Lattice Boltzmann Method

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Introduction

In recent years, the Lattice Boltzmann Method (LBM) has been demonstrated as an interesting alternative for the simulation of complex fluid flows. Unlike finite elements models (FEM) based on a discretization of macroscopic continuum equations, the LBM is derived from microscopic models and mesoscopic kinetic equations [1]. As an increasing number of studies use acoustic standing waves for microfluidics applications [2], the LBM constitutes an efficient alternative to simulate complex interactions between fluid flows and acoustic waves. The main advantages include clear physical pictures, easy implementation of boundary conditions and a fully parallel algorithm. This paper shows results towards the implementation of LBM for the creation of acoustic standing waves in straight microchannels.

Simulation models

We implemented two models using a conventional LBM code (similar to [3]) solved with Matlab™. A 1D-model represents a channel filled with liquid incorporating a piezo actuator on one wall and a reflecting wall on the opposite side. As standing waves are a one-dimensional problem, we have defined a very narrow lattice of 3 per 1000 nodes for the channel model. Acoustic waves were introduced into the model at $x = 1$ by point-source nodes with the velocity vectors fixed to zero and the density defined as:

$$\rho(t) = \rho_o + A\sin(\omega t) \qquad (1)$$

The angular frequency ω was fixed at the wavelength of $\lambda = 2 \cdot (L - 1)$ which is the condition for the establishment of a standing wave with one central pressure node. On the opposite wall at $x = L$, we have implemented a no-slip boundary according to the Zouh & He method [4]. Following the 1D-simulation, we implemented a 2D-model with a wave generation only at the central part of one channel wall (100 lattice-nodes wide). The 2D-model consists of a wider matrix (500 x 200 nodes) with the same boundary conditions as the previous model.

Results

From the 1D-model, one can observe the generation of a transient wave until a standing wave was established (Fig. 1). After about 6000 simulation cycles, a nearly perfect standing wave pattern was obtained according to the stable density reached by the central node at the middle of the channel (Fig. 2). In the 2D-model, the wave is confined around the point-source nodes at the beginning of the simulation (Fig. 3:a). Further away in the simulation, a standing wave is formed along the length of the channel even if the pressure source is confined to the channel center. (Fig. 3,b).

Conclusion

We have successfully implemented two simulation models using the LBM. The possibility to obtain pressure standing waves showing the origin of their formation was demonstrated in a 1D-model. In a 2D-model, the propagation of a wave from the central part of the channel showed resulting planar standing waves that are continuous along the full length of the channel. The present results demonstrate that the LBM is a promising method for simulations of complicated acoustic phenomena towards the development of efficient acoustic microfluidic devices.

Tuning the ultrasonic particle manipulation performance by the angle and position of a wedge transducer

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Introduction

Transducers combined with wedges can be used for efficient coupling of ultrasound into chip-based standing-wave manipulation devices [1]. However, the chip – wedge transducer systems developed previously in our group have generally complex geometric shapes, generating complex 3D acoustic resonances. Consequently, it is very difficult to accurately predict the resonances by theoretical modeling, as well as to fabricate transducers and chips with matching resonance frequencies even if the optimum design criteria were known. To overcome this problem, we here present a novel transducer design approach based on a continuously tunable wedge angle. We investigate the dependence of the trapping pattern of 5 µm beads on the wedge angle, as well as on the transducer position on the chip surface.

Experiment

In the first experiment, an in-house built tunable-angle transducer (see Fig. 1a) is operated around 2.13 MHz. The coupling angle θ of the ultrasonic beam can be continuously selected between 17 and 73˚ relative the normal to the chip surface. In a second experiment, a fixed-angle transducer operated around 2.4 MHz was moved over the upper surface of the chip as indicated in Fig. 1b. Here, four different transducer positions were investigated, marked with 1, 2, 3 and 4 in Fig. 1b. In both experiments, 5 µm polystyrene beads were used for visualizing the trapping pattern.

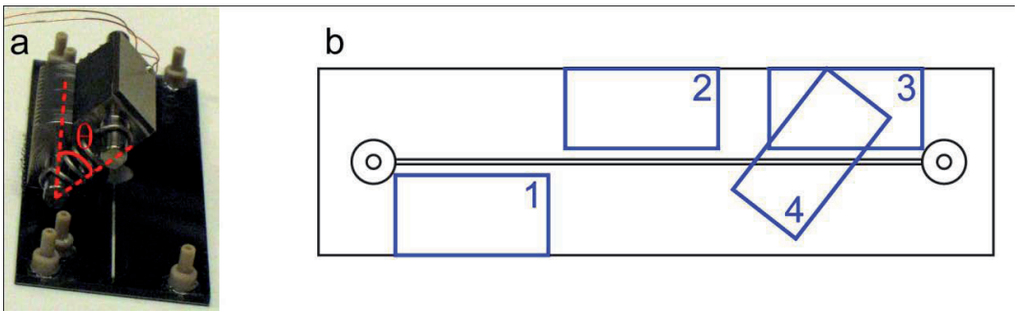

Fig 1. a) The tunable-angle wedge transducer, and the microfluidic chip. b) The investigated positions of the fixed-angle transducer.

Purification and disinfection of water with high frequency ultrasound

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Introduction

Water treatment is of increasing importance in today's world, as there is an increasing demand for safe drinking water, while an increasing number of people cope with water scarcity due to climate changes. Secondly, there is need of water of varying quality for industrial applications and agriculture. Last but not least, water that has been used (by humans) for some application, needs to be treated before discharge into the environment.

For preparation of potable water from ground water, suspended particles need to be removed. This can be done in several ways, mainly by using physical methods, like filters. The disadvantage of these filters is that they usually becomes clogged at some point in time, moreover filters are polluted by bacterial growth. It would therefore be very beneficial if a filter exists, which does not use a medium, but which filters on the basis of another principle instead of particle size combined with a porous structure.

A method which seems viable in this respect is acoustical filtering. Acoustic filtration uses ultrasound to concentrate tiny particles in a fluid. The method proved to work in water applying small flows, as shown by various authors [1-5]. An important research question is how to design an energy efficient acoustic filter in such a way that scaling up to industrial dimensions is possible and in what frequency range such a filter should be operated. It is expected that operating frequency and energy efficiency of ultrasound equipment for water treatment can be improved by applying optimised resonant structures and adequate control strategies.

The first and main objective of our research is to find a efficient method of filtering particles from 10 up to 500 micrometer. A secondary objective could be to disinfect water using ultrasonic waves.

Approach

In this research a combination of numerical simulations and experiments is applied. First of all an existing separator was modelled in 1D with Matlab software. In addition to this, a sensitivity analysis was performed with respect to particle trajectory and expected separation

Acoustic sorting of encapsulated cells

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Introduction
Polymer microencapsulation of cells has attracted researchers from various research fields such as tissue engineering and cell therapy, by isolating, functionalizing and culturing specific cells in three-dimensional (3D) way. So far, many studies suggested a microfluidic method of droplet-generation which can produce funtional microbeads, spheres and/or capsules with different purposes. However, it is not easy to obtain uniform dimension or number of cells inside [1-3].

Experiment
In this manuscript, we introduce a novel microfludic method to sort microbeads and capsules by standing surface acoustic wave (SSAW). The basic working principle is similar to those in prior studies for particle separation [4], with modified and optimized structures and dimensions of the channels and interdigitalized transducer (IDT) electrodes for microbeads/capsules sorting.

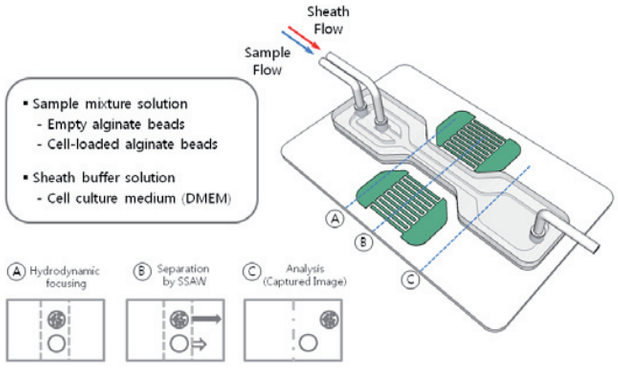

Figure 1. Schematic of working principle of microfluidic device on acoustic separation of microbeads

The device was fabricated through plasma bonding process on a PDMS microfluidic channel structured for hydrodynamic focusing and inter-digitalized transducer (IDT) deposited on lithium niobate (LN) wafer for SSAW generation. To demonstrate the sorting capability of this device, we prepared several types of alginate beads (empty; cell-loaded; cell-loaded with different amount) by microfluidic droplet-generating device and then normalized the displacement of under SSAW.

Results
As illustrated in Fig. 1, a mixture of different types of beads was injected into center channel, and then squeezed by sheath (DMEM) flow. The focused beads experienced acoustic radiation force, and moved perpendicular to the main flow dependent on the different material properties of microbeads and number/properties of cells inside. The sorting process was monitored and recorded by a camscope to analyze efficiency and performance. We found that various types of alginate beads can be completely

Acoustophoretic enrichment of sub-micrometer particles

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Introduction

The ability to fast and easy separate and enrich sub-micrometer particles e. i. bacteria, has become increasingly important in health care, food and environment applications. [1] Using acoustophoresis particles can be separated based on size and density. [2] When operating at a frequency of 2 MHz a line can be seen around 1µm in particle diameter under which particles can no longer be focused. One way to be able to focus smaller particles is to increase the frequency. In this study a single node 6 MHz chip was used to enrich 1µm and 600 nm diameter polystyrene particles.

Experiment

The chip was constructed in silicon with a bonded glass surface. <110> oriented silicon was used to be able to perform anisotropic etching and thus get a channel deeper than wide, with the dimensions 125*220 µm, thus allowing higher throughput. The width of the channel corresponds to a single node standing wave with the frequency 6 MHz. (See figure 1.)

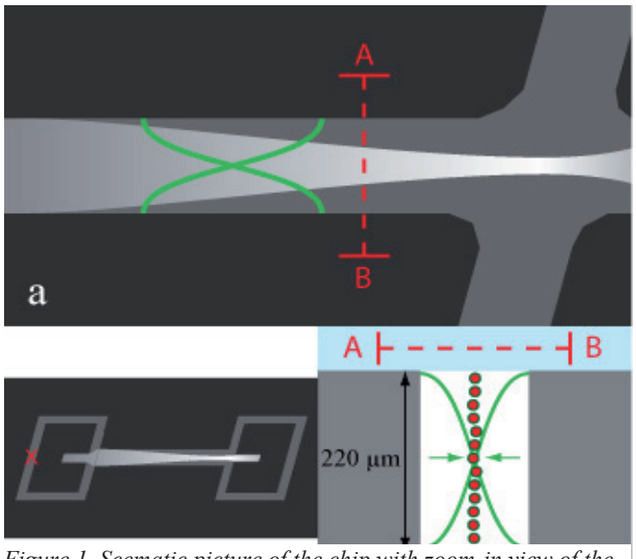

Figure 1. Scematic picture of the chip with zoom-in view of the outlets with the standing wave and a focusing stream of particles (a), a zoom-out view (b) and cross-section shownig the standing wave and the focusing particles (c).

0.1 weight % polystyrene particles with diameter 1 µm and 600 nm in MQ water was introduced into the chip through a single inlet, acoustically focused in a single node in the channel center and extracted though the middle outlet while excess water was extracted through the side outlets. The optimal frequency was found to be 6.13MHz and throughout the whole experiment the voltage was kept at 50 $V_{p\text{-}p}$. The particle distribution was determined by counting extracted fractions in a coulter counter.

Results

The outlet distribution for the 1 µm particles for a variety of split ratios is shown in figure 2. The numbers show a clear trend with over 88 % of the particles leaving then chip trough the middle outlet for all cases. Figure 3 show the outlet distribution of 600 nm particles. The trend is

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