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Deterministic separation of cancer cells from blood at 10 mL/min

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Circulating tumor cells (CTCs) and circulating clusters of cancer and stromal cells have been identified in the blood of patients with malignant cancer and can be used as a diagnostic for disease severity, assess the efficacy of different treatment strategies and possibly determine the eventual location of metastatic invasions for possible treatment. There is thus a critical need to isolate, propagate and characterize viable CTCs and clusters of cancer cells with their associated stroma cells. Here, we present a microfluidic device for mL/min flow rate, continuous-flow capture of viable CTCs from blood using deterministic lateral displacement (DLD) arrays. We show here that a DLD array device can isolate CTCs from blood with capture efficiency greater than 85% CTCs at volumetric flow rates of up to 10 mL/min with no effect on cell viability. *Copyright 2012 Author(s). This article is distributed under a Creative Commons Attribution 3.0 Unported License*. [http://dx.doi.org/10.1063/1.4758131]

Circulating tumor cells (CTCs) and circulating clusters of cancer and stromal cells (the latter forming the "seed and soil" hypothesis of metastasis[1]) have been identified in the blood of patients with malignant cancer and can be used as a diagnostic for disease severity, assess the efficacy of different treatment strategies[2,3] and determine the eventual location of metastasic invasions for possible treatment.[4] There is thus a critical need to isolate, propagate and characterize viable CTCs and clusters to expand on understanding of the biology of metastasis,[5,6] enable new methods for studying the emergence of drug resistance,[7] and aid in the development personalized treatments.[8,9] Because of the rarity of CTCs (1-1000 cells/mL)[3] compared to other cells in the blood and the large variation in their morphology,[10] large volumes have to be processed to deliver a sufficient number for statistical detection and detailed characterization of the heterogeneity.

Much of the recent work in this area has focused on capturing cells on epithelial antibody-coated surfaces in microfluidic flow channels,[11,12] but these devices must be operated slowly ($\sim$ mL/hr) to maintain capture efficiency and viable cell recover is difficult because cells are strongly bound to the surface once they attach. CTCs (diameters 15–30 μm) are on average larger than other cells in blood (2–15 μm),[13] giving the possibility of enrichment with a size-based separation technique. Circulating clusters of cancer and stromal cells are even larger, up to 200 μm in diameter.[6] Several methods involving membrane based filtration[14–16] or devices relying on inertial effects[17,18] have been developed to exploit this size difference. Because of the need for a large range of sizes, clogging may be an issue with membrane-based technology, while the throughput of inertial techniques is limited by the heavy dilutions that are necessary to maintain separation resolution.

Deterministic lateral displacement (DLD) is a microfluidic size-based particle sorting method with excellent size selectivity, adaptability to sorting multiple particle sizes,[19] and dynamic control

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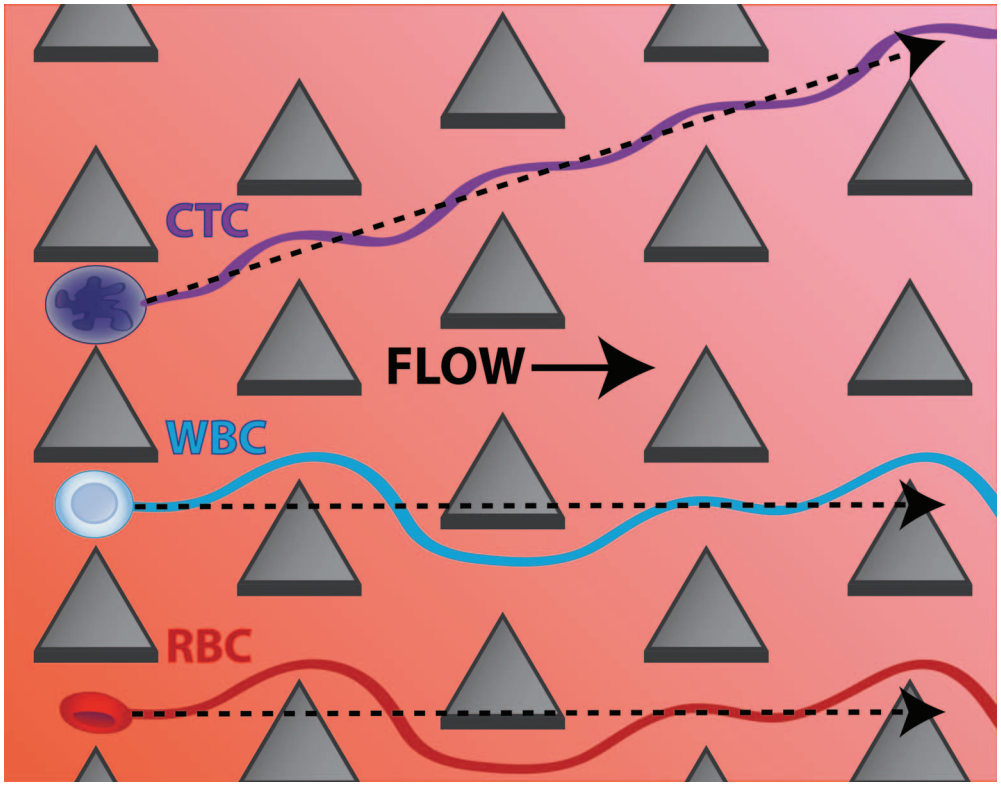

FIG. 1. **CTC Separation in DLD Array.** Circulating tumor cells can be separated from other cells in the blood after passing through a deterministic lateral displacement (DLD) array. An array of microfabricated posts direct particles above a critical size (CTCs) at an angle with respect to the fluid flow direction while small particles (WBCs and RBCs) follow the fluid flow direction. Using such an array, CTCs can be concentrated from an input stream and collected at the end of the array.

of critical particle sizes.[20] It has been demonstrated under a broad range of operating conditions, sorting particles from 100 nm[21] to 30 um.[22] In the method, as shown in Fig. 1, an array of microposts direct particles above a critical size (CTCs) at an angle with respect to the fluid flow direction so that the larger particles become concentrated at one sidewall of the array. Using such an array, large particles can be concentrated from an input stream and harvested at the end of the array by collecting the output fluid stream separate from the rest of the fluid leaving the array. However, it has never been demonstrated that a deterministic lateral displacement array can process the volumes needed for selection of rare circulating cells in whole blood in times of the order of minutes. Here, we present a DLD array device for mL/min flow rate, continuous-flow capture of viable CTCs from blood using deterministic lateral displacement arrays. We show that a DLD array can be designed such that it will isolate CTCs from blood with efficiency greater than 85% CTCs over a large range in sizes from milliliter-volume blood samples in minutes, with no effect on cell viability so that further culturing and analysis of the cells can be carried out.

The layout of the concentrator device used for these experiments is shown in Fig. 2. It is composed of a 2.5 mm wide by 25 mm long flow chamber filled with a mirrored array of 58 μm triangular posts with 42 μm gaps between posts. One axis of the array is tilted at an angle of 1/20 radians with respect to the direction of fluid flow, which should give a critical size of 7 microns. A unique aspect of the deterministic bump array present here is that triangular posts are used rather than circular posts, which should provide for increased throughput and less clogging compared to circular posts.[23] Inglis' wall design is used to ensure flow uniformity near channel walls where particles are concentrated.[24] According the operating principles of DLD, large particles above a designed critical size (e.g. the cancer cells) will flow along this tilted axis of the array, while carrying fluid and small particles flow in the horizontal direction defined by the array sidewalls. The top

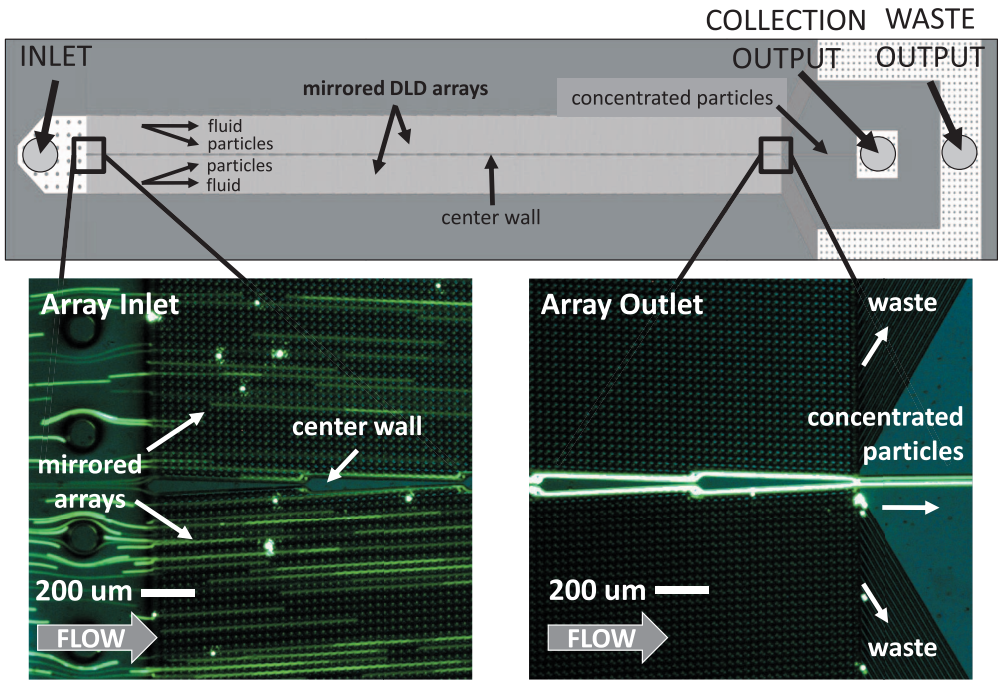

FIG. 2. **High Throughput Device**. A concentrator deterministic lateral displacement array with one input and two outputs (collection and waste) is used. Fluorescent micrographs of the trajectories of 3.1 μm beads in a demonstration array with with 12 μm gaps, 18 μm triangular posts, 1/20 array tilt and 2.0 μm critical size highlight device function. Using a mirrored design, large particles dispersed in the inlet are focused against the central channel wall, where they can be collected at the narrow collection output while other particles enter the waste outlet.

section in the mirrored array directs large particles downward towards the central wall while the bottom section directs them upward. This allows the device width to be doubled without increasing the array length, which must typically be scaled with the array width divided by the array tilt to ensure all large particles are displaced to the central wall. A fluid containing targeted particles is brought in from a single inlet and flowed through the array. While traversing the array, large particles are concentrated towards the center of the flow chamber and collected at a narrow collection output while all other fluid is directed to a waste outlet.

Devices are etched into a silicon wafer to a depth of 60 μm and sealed with a PDMS-coated glass cover slide. Three devices are tiled in plane on each silicon layer and external connections are made by mating an acrylic manifold to through-wafer holes etched into device at inlet and outlet ports. The manifold is spring-loaded to withstand applied pressures up to 4.7 atmospheres without leaking. Manifold, devices, and tubing are first wet with a running buffer of 1× PBS, 1% BSA, and 1 mM EDTA. 5 mL of fluid was processed in each experiment at various flow rates that were controlled by a syringe pump. The pressure drop across the device was measured at 4 atmospheres at 10 mL/min flow rate for buffer and scaled linearly with flow rate.

Experiments were performed with approximately 10^7 cancer cells from breast or prostate cell lines in 1 mL of growth medium spiked into buffer or blood diluted with buffer. Cells are cultured using standard methods and used immediately after passage. Premalignant (MCF10A) and malignant (MDAMB231) epithelial breast cells contained a GFP or RFP construct that allowed them to be easily differentiated from other non-fluorescent cells in solution.

Device functionality with cancer cells was confirmed by observing the trajectories of MDAMB231 breast cells expressing GFP with epifluorescence microscopy at a flow rate of 100 $\mu L/min$ (Fig. 3). Using cells in buffer, we can clearly see that cells enter the device broadly distributed from a single inlet as they enter the array and are focused against the central wall and directed into the collection output at the end of the array.

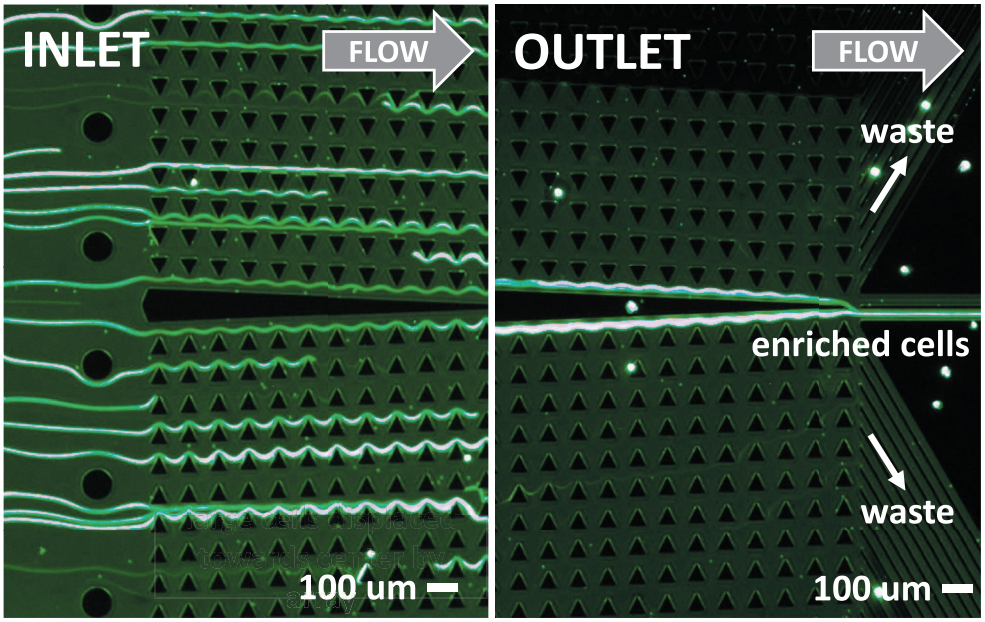

FIG. 3. **CTC enrichment in DLD array**. Trajectories of GFP-expressing MDAMB231 breast cells in an array with 42 μm gaps, 1/20 array tilt, and 7 μm critical particle size at device inlet and outlet at flow rate of 500 $\mu L/min$. Cells are evenly dispersed at inlet but are concentrated against central wall at outlet and directed to narrow collection outlet.

We confirmed the function of the DLD array at high flow rate by processing MCF10A breast cells spiked in the wetting solution at 10 mL/min. 1 mL of growth medium with $\sim 10^7$ cells was diluted to 10 mL with the wetting solution and 5 mL of sample was processed in 30 s. Concentrations and size distribution of cells were measured for input, collection, and waste solutions using a Coulter counter. From an initial concentration of 3.75×10^6cells/mL with average size 19.5 μm, the collection output had a concentration of 2.1×10^7 cells/mL with average size 19.5 μm while waste output had a concentration of 3.7×10^5 cells/mL with average size 19.0 μm. Factoring in the volume of each output ($V_c = 0.8$ mL, $V_w = 4.2$ mL), we can calculate the capture efficiency as the ratio of cells in the collection output to the total number of cells in both outputs $\frac{\#\, collected\ cells}{total\ cells} = \frac{C_c V_c}{C_c V_c + C_w V_w}$. Using this definition, 91% of the targeted cells were collected by the device.

Experiments were then performed with cancer cells spiked into blood to assess the potential for separating CTCs from blood. Blood was supplied by donors from the Interstate Blood Bank (Memphis, TN), shipped overnight and used within one day. Since blood has a higher viscosity than water, a dilution between 5 and 20 times was performed to ensure that the device operated below the manifold pressure limit. In a first experiment, 500 μL of blood was added to 1 mL of growth medium with $\sim 10^7$ malignant MDAMB231 breast cells and then diluted to a total volume of 10 mL with the wetting solution. 5 mL of this mixture was processed at 10 ml/min flow rate and the output solutions were analyzed with a flow cytometer. MDAMB231 cells used expressed a red fluorescent protein dTomato that was used to distinguish them from other cells in the mixture which had no fluorescence. 50,000 total cells were analyzed from the input, collection output and waste output. In each run and the number of large, fluorescent cells were enumerated from multivariable analysis. 4.9% of the input cells were cancer cells, while 16.7% of collection output cells and 0.6% of waste output cells were cancer cells. If we assume that the concentrations of other cells remain constant, we can normalize these percentages to the ratio of cancer cells to other cells in each solution and get 0.05, 0.2 and 0.006 for input, collection and waste solutions respectively and a capture efficiency in these conditions of 86%.

For 10 mL/min flow rate, the fluid velocities in the device range from near zero at the post edges to $\sim$ 1.5 m/s in the middle of a gap between posts. This subjects the cells to an high shear stress that might damage them. The strength of shear forces in a flow are characterized by a shear rate

TABLE I. **Viability Experiment Summary.** V_{max} defined as maximum velocity across a gap extracted from numerical simulation, shear rate $\partial v_x/\partial y$ estimated as maximum velocity in the gap divided by half the 42 μm gap. Array length is 25 mm.

Flow Rate (mL/min)	Cell Type	Viability	Cells Counted	V_{max} (cm/s)	Shear Rate (s^{-1})	Transit Time (s)
0.1	MDAMB231	0.95	61	1.5	750	7.0
0.1	PC3	0.96	28	1.5	750	7.0
0.5	MCF10A	0.96	51	7.5	3750	1.4
5.0	MCF10A	1.00	123	75	37500	0.14
10.0	MCF10A	0.99	125	150	75000	0.07

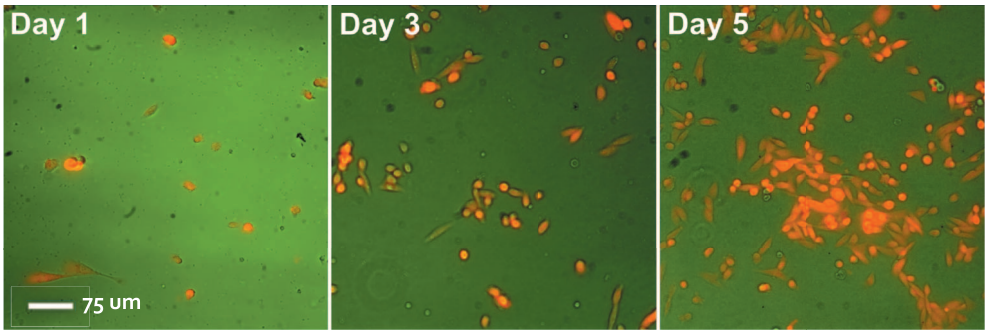

FIG. 4. **MDAMB231 cells are unaffected by passage through the DLD array.** Time series of cancer cells (red) spiked in blood shows proliferation in a culture flask after flow through device at 5 mL/min flow rate.

defined as $\partial v_x/\partial y$, where x is the flow direction and y is the direction across a gap. Cell viability was measured by application of 0.4%w Trypan blue in saline at a 10:1 dilution to input and output samples, vortexing, incubating for 5 min at room temperature, and examining the resulting solution with a haemocytometer.

Passing through the DLD array had no effect on cell viability in several different cell lines suspended in the running solution across a wide range of flow rates. In experiments, summarized in Table I with three epithelial cell lines (MDAMB231, PC3, MCF10A) at various flow rates from 0.1 to 10 mL/min, greater than 95% of the cells survived in each experiment. PC3 are malignant prostate cancer cells. While the shear rates in these devices greatly surpasses physiological levels (1–2000 s^{-1}[25]), we speculate that the reduced exposure time at the higher flow rates limits cellular damage. At 10 mL/min, cells traverse the entire 25 mm array in about 70 ms. This observation matches with observations of white blood cells in a rotary pump that suggest shear-induced damage is an integral effect over time.[26] That is, cell vitality decreases the longer cells are exposed to high shear. Experiments done by Di Carlo *et al.*[27] to examine gene expression in cells exposed to high shear showed no significant changes. As a result, we do not expect shear-induced cell death or alteration to be a significant limitation in the operation of these devices.

We further assessed cell vitality by culturing cells after they had passed through a device. Approximately 10^6 MDAMB231 cells were spiked into whole blood diluted 10:1 and 5 mL of this solution was driven through a single layer device at 5 mL/min. Red blood cells in the output from the collection channel were lysed with 1× RBC lysis buffer (eBiosciences), washed with the running solution and 500 μL of the resulting solution was placed into a 25 cm^2 culture flask with 3 mL of growth media. Shown in Fig. 4, the red fluorescent MDAMB231 cells attach to the surface and proliferate. While there is a small background of white blood cells and unlysed red blood cells, the cancer cells seem unaffected.

The 10 mL/min flow rates presented constitute a 50-fold improvement in flow rate over the herringbone-CTC chip used by Stott *et al.*,[12] a 10^5-fold improvement over the 10 s of $\mu L/min$ historical flow rates in DLD arrays[22,28] and a 100-fold improvement over more recent efforts by

Inglis[29] and are the fastest reported operation of these devices. With internal velocities exceeding 1 m/s and Reynolds number Re $>$ 40, we confirm that the size-based separation functionality is preserved even outside the low Reynolds number flow regime.

However, the number density concentration of about $\times 4$ needs to be greatly improved for rare cell capture. There are several improvements that can be made to increase this number. (1) A second buffer input should be added to direct cells into a clean solution free from background cells in blood. Because this device has only a single inlet, there will always be a background of blood cells in the collection output that may interfere with attempts to propagate cells or analyze them. There are several examples in the literature that have demonstrated that cells in blood can be isolated literally free from background cells using DLD arrays with a buffer input.[22,28,29] (2) Although a second clear buffer channel would decrease the background of non-sorted cells greatly, it would not increase the absolute concentration of sorted cells in the output stream.

The absolute concentration of sorted cells is the product of the initial concentration times the efficiency of isolation, presently about 91%, times the ratio of the volume/sec of fluid flow in the waste stream to the flow in the collection stream. We have 34 channels (17 on each side) feeding a common collection channel, but the collection channel is 1.6 times the width w_{array} of the channels in the array, so it would appear that concentration should be 34/1.6 = ($\times 21$), but this ignores the dependence of the flow on the dimensions of the channels. The actual computation of array conductance is complex, but we can guess what it is by roughly using the thin slit approximation[30] for the conductance/length $\mathcal{C}$ of a thin slit:

$$\mathcal{C} = \frac{h^3 w}{12\eta} = \frac{Q}{G} \tag{1}$$

where η is the fluid viscosity, h is the height of the slit, w is the width ($h \ll w$ for a thin slit), Q is the volume/sec of flow and $G = \frac{dP}{dz}$, the pressure gradient. Since the channel depth in our case is only twice the width, this overestimates the difference in conductance between the product outlet channels and the waste outlet channels. Empirically, a change of the exponents from $h^3 w$ to $h^{2.4} w^{1.6}$ yields a concentration of $\times 6$ for our device. Since the efficiency of capture in our device is quite high, the simple expedient of feeding the output from the collection channel into another array would increase the concentration exponentially. A 200 mm wafer could easily accommodate 6 of these devices in series, and also allow simply scaling the device up in size with more side channels combined with a narrower collection channel. We think a concentration enhancement of 10^5 at high flow rates with very low background is easily within reach.

In summary, these experiments showed that high efficiency capture of cancer cells from blood is possible at flow rates of 10 ml/m with no effect on cell vitality by using deterministic lateral displacement arrays at flow rates much higher than used in previous microfluidic devices. Operating devices in parallel by tiling more devices in plane and vertically stacking devices could easily enable flow rates an order of magnitude larger. These results open up the possibility of rapid collection of viable circulating tumor cells and clusters of cancer and stromal cells from the blood of patients with metastatic disease. Capturing and propagating viable cells from patients could allow *in vitro* chemotherapeutic treatments to be studied in vitro to determine which may be the most effective in vivo and allow better study of the development of drug resistance.

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